THE RELATIONSHIP BETWEEN GAIT BIOMECHANICS AND BODY MASS
INDEX
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Walking gait is one of the basic components of human movements. With the

rising obesity epidemic and implied health complications, it is pertinent to examine the

relationship between body mass index and the joint torques and powers at the hip, knee,

and ankle joints required to produce gate. The purpose of this study was to identify the

relationships between ground reaction forces, joint torques, and joint powers at the hip,

knee, and ankle and BMI in a group of people with BMIs ranging from 18-44 kg/m^2,

while walking at a self-selected normal speed and at a safe self-selected maximum speed.

We proposed a direct relationship between BMI and ankle torques and powers.

Additionally, we proposed that hip and knee joint torques and powers would have a direct

relationship to BMI up until approximately 35 kg/m^2 at which point individuals would

begin to modulate their hip and knee joint torques and powers to mimic those used by

healthy individuals.

Gait biomechanics were collected for twenty-one healthy adults with BMI

between 18 kg/m^2 and 44 kg/m^2 using an eight-camera Qualysis motion capture system

and evaluated using Visual 3D. Each subject was tested while walking at a normal speed

and a safe-maximum speed. The data showed a statistically significant relationship

between maximum hip extensor torque at both a normal speed and a safe-maximum

speed. No significant relationships were found between knee torques or powers and BMI.

Maximum ankle plantarflexor torques and maximum positive ankle powers were found to

be significantly correlated with BMI for both speeds. Additionally, first and second maximum ground reaction forces and maximum braking and propulsive forces were found to be significantly correlated with BMI at both speeds.

We were unable to fully evaluate the specific hypothesis due to an inadequate sample size however, our results suggest that further investigation may support the hypothesis. Additionally, our subject sample was largely skewed towards females. We intend to continue subject recruitment in order to wholly test our original hypothesis as well as evaluate possible differences between the gait biomechanics of Class III obese males and females.

The Relationship Between Gait Biomechanics and Body Mass Index

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Table of Contents

List of Tables

List of Figures

Chapter 1: Introduction

Walking gait is a basic component of human movement. Through locomotion, humans are able to move their body mass through space at a given velocity. Walking is characterized by the cycle of moving one lower limb through space while supporting oneself on the contralateral limb followed by a transfer of weight in order to swing the second limb (Murray, 1967). A stride cycle is characterized by the time between two successive heel strikes of the right or left foot (Murray, 1967). One stride cycle is comprised of a stance phase, when the foot is in contact with the floor, and a swing phase when the foot is moving through the air (Murray, 1967). Stride length can be defined as the distance between successive points of heel strike for a single foot (Murray, 1967). Step length is measured as the distance between the points of contact of the heel with the ground of the right foot followed by the point of contact of the left foot (Murray, 1967).

Changes in gait patterns induced by obesity may be a factor for the development of many musculoskeletal disorders. For example, gait differences are exacerbated in obese individuals with low back pain compared to lean individuals with back pain. These subjects are seen to have reduced stability, prolonged stance, lower velocity, and shorter step length compared to both obese individuals with no back pain and lean individuals with back pain (Cimolin et al., 2011). Load carriage has a vast impact on the health of the human foot as well. Heel pad thickness and heel pad compressibility index are significantly greater in obese individuals with heel pain than normal weight individuals (Prichasuk et al., 1994).

Increased body mass index is strongly correlated with an increased incident rate of knee osteoarthritis (Reijman et al., 2007). Individuals with knee osteoarthritis have

lower mean sagittal knee angular velocity as well as lower maximum sagittal plane knee extension velocity than healthy individuals. Affected individuals also have lower mean sagittal knee extension values for each leg with a greater decrease in the affected leg, less sagittal knee range of motion in the affected leg, and lower mean sagittal hip angular velocity than healthy individuals (Messier et al., 1992). The occurrence rate of knee osteoarthritis is proportional to the degree to which an individual is overweight (Tukker et al., 2008). A particularly strong association exists between obesity and osteoarthritis of the knee (Tukker et al., 2008). For each unit increase in BMI, individuals are 8% more likely to suffer from osteoarthritis or chronic knee pain (Tukker et al., 2008). Women exhibit a higher association between being overweight and suffering from knee osteoarthritis while overweight men have a higher affinity for hip osteoarthritis than women (Tukker et al., 2008). Being overweight not only increases the risk of developing osteoarthritis, but also increases the risk of decreased mobility in the general population as well as within the portion of the population suffering from osteoarthritis (Tukker et al., 2008).

Joint torques are frequently used to describe human gait. Torques can be considered as the effect of forces exerted by a muscle crossing a joint (Vaughan, 1996). Magnitude and direction of joint torque can be estimated by examining the magnitude and direction of ground reaction forces (GRF) (Vaughan, 1996). This is done by multiplying the GRF and the perpendicular distance from its line of action to the joint center (Vaughan, 1996). Human gait is often divided into the phases of heel strike, foot flat, midstance, heel off, and toe off (Vaughan, 1996). In gait patterns of healthy, lean individuals, GRF passes anterior to the hip joint and posterior to the knee and ankle joints

just after heel strike leading to extensor torques at the hip and knee joints and dorsiflexor ankle torque (Vaughan, 1996). When the foot is flat on the ground, GRF is anterior to the hip, through the knee joint center, and anterior to the ankle joint center resulting in a large hip extensor torque, zero torque at the knee, and plantarflexor torque at the ankle (Vaughan, 1996). At midstance, GRF is reduced in magnitude and passes through the hip and anterior to the knee and ankle (Vaughan, 1996). During midstance, the hip torque is zero, knee torque is flexor, and a large lever arm at the ankle results in a large plantarflexor torque (Vaughan, 1996). When the heel leaves the ground, GRF are increased to a magnitude greater than body weight and passes posterior to the hip, posterior to the knee, and anterior to the ankle. At heel off the hip has a flexor torque, the knee has an extensor torque, and the ankle has a large plantarflexor torque (Vaughan, 1996). GRF magnitude is again decreased just before toe off and located posterior to the hip and knee and anterior to the ankle (Vaughan, 1996). Joint torques at toe off are in the same direction as heel off but smaller in magnitude (Vaughan, 1996).

Obesity produces numerous variations in joint biomechanics during gait. General categorizations define moderate obesity as having a BMI between 30 kg/m^2 and 40 kg/m^2 and Class III obesity as having a BMI above 40 kg/m^2. Class III obese individuals walk with a larger support base accompanied by an increased forefoot abduction angle, shortened phases of swing and single support, increased double support, and a reduction in stride length (de Souza et al., 2005). Stance duration appears to be directly proportional to BMI with an extended duration as compared to lean individuals (Blaszczky et al., 2011). Ground reaction forces and torques are also closely related to

BMI. Obese individuals experience larger ground reaction forces and joint torques at the hip, knee, and ankle joints than normal weight individuals (Browning et al., 2007).

Moderately obese individuals, characterized by Browning, et al. as having a mean BMI of 37 kg/m^2 for females and 34.1 kg/m^2 for males, walked with similar stride kinematics but altered kinetics as lean individuals (Browning et al., 2007). Stride length and stride frequency did not differ between moderately obese individuals and normal weight individuals (Browning et al., 2007). In these persons, ground reaction forces were larger in every direction. Moderately obese individuals had approximately 60% greater ground reaction forces than normal weight individuals (Browning et al., 2007). Joint torques of moderately obese individuals were proportionately larger at the hip, knee, and ankle joints compared to normal weight individuals (Browning et al., 2007). There is a mechanical relationship in which an increase in body mass leads to a concurrent increase in forces and torques (Browning et al., 2007; Sheehan et al., 2013). Moderately obese individuals show little adaptation to increased mass.

Whereas there exists a linear relationship between gait biomechanics and level of obesity in lean to moderately obese individuals, there appears to be a nonlinear change in the gait biomechanics of Class III obese individuals, characterized as having a mean BMI of 42.3 kg/m^2 (DeVita et al., 2003). Class III obese individuals walked more slowly with shorter step lengths and exhibited significantly larger ground reaction forces than lean individuals. When walking at similar velocities, Class III obese individuals had a 5% shorter relative swing time and 3% longer relative stance time than lean individuals (DeVita et al., 2003). These obese individuals walked more erect with ~5º more hip extension throughout stance, ~8º less knee flexion in early stance, and ~4º less knee

4

flexion throughout stance compared to lean individuals (DeVita et al., 2003). In addition, Class III obese individuals were ~6º more plantarflexed throughout the stance phase and ~7º more plantarflexed at toe off than lean individuals (DeVita et al., 2003). Most importantly, joint torque and power patterns of Class III obese individuals were similar to those of normal weight individuals at the hip and knee and only varied in magnitude at the ankle joint (DeVita et al., 2003). This shift in hip and knee joint torques and power potentially creates an inverse relationship between joint torque and power and BMI in Class III obese individuals. Class III obese individuals used 89% more plantarflexor angular impulse which produced 61% more work at the ankle joint compared to the normal weight individuals (DeVita et al., 2003). Class III obese individuals adapted to their excessive mass by using more torque at the ankle to propel them upward and forward. The use of increased ankle torques and reduced hip and knee torques suggests the existence of a unique neuromuscular adaptation to obesity in Class III obese adults (DeVita et al., 2003).

In summary, overweight to moderately obese individuals walk with similar kinematics as lean individuals, larger ground reaction forces, and joint torques proportional to their mass. There is a linear relationship between vertical ground reaction forces and sagittal plane joint torques with BMI; as BMI increases, ground reaction forces and joint torques increase. Obesity elicits mechanical adaptations in moderately obese individuals. The different adaptations seen in the Class III individuals suggest that these individuals make behavioral adaptations that enable them to alter the relationship between mass and hip and knee biomechanics in addition to mass driven adaptations at the ankle. Based on the reported gait biomechanics, it appears that these behavioral

adaptations may appear at BMI value between ~35 and ~45 kg/m^2. At this point, hip and knee biomechanics may become inversely proportional to BMI. In short, Class III obesity prompts behavioral changes in gait that are qualitatively different than obesity-induced changes. Alterations to gait patterns in obese individuals may contribute to an increased occurrence of disability.

Hypothesis

It was hypothesized that there is a critical magnitude of obesity, as measured with Body Mass Index, above which humans walk with an inverse relationship between BMI and hip and knee joint torque and power. The expected results are: that BMI will be: 1) directly related to GRFs and ankle joint torque and power and 2) directly related to hip and knee joint torques and powers from BMI 20 to ~35 kg/m2 and indirectly related at higher BMI values.

Purpose

The purpose of this study is to identify the relationships between ground reaction forces, joint torques, and joint powers at the hip, knee, and ankle and BMI in a group of people with BMIs ranging from 18-44 kg/m^2, while walking at a self-selected normal speed and at a safe self-selected maximum speed.

Delimitations

1. With the exception of obesity, all subjects will be apparently healthy, mobile adults with no history of neuromuscular, musculoskeletal, physiological, or neurological disability.

2. Individuals with previous surgery on the lower limbs will be excluded.

3. BMI will range from 20-50 kg/m^2.

4. Age will range from 18-60 years.

5. Analysis will be performed on the right lower extremity of all subjects.

Limitations

1. Accuracy of health surveys

2. Accuracy of marker placement and 3D Motion Capture System

3. Accuracy of AMTI Force Plate

4. Subjects may adjust their normal gait patterns after marker placement and in an unfamiliar environment.

Chapter 2: Review of Literature

The primary intent of this thesis is to determine the relationship between ground reaction forces, joint torques, and joint powers at the hip, knee, and ankle and BMI in individuals with BMI's ranging from 20-50 kg/m^2. More specifically, it will attempt to determine at what level of obesity individuals begin to adjust their gait patterns as an adaptive technique to enable the transport of greater mass loads. This chapter will review literature with a focus on 1) the influence of obesity on health parameters, 2) gait characteristics of normal, lean adults and obese adults 3) techniques used to measure gait characteristics.

Influences of Obesity on Health Parameters

Obesity is a rising epidemic in developed countries as defined by having excess body fat. Body mass is commonly measured by mass in kilograms divided by height in square meters, otherwise known as body mass index (BMI) (Grundy et al., 2004). The Center for Disease Control classifies BMI under 18.5 kg/m^2 as underweight, 18.5-24.9 kg/m^2 as normal weight, 25.0-29.9 kg/m^2 as overweight, and 30.0 kg/m^2 and above as obese (CDC, 2011). Obesity has been further categorized based on incurred health risk. Body mass index ranging from 30-34.9 kg/m^2 is considered class I obesity, 35-39.9 kg/m^2 is class II, and over 40 kg/m^2 is class III (Flegal et al., 1998; Katzmarzyk et al., 2006). As an individual moves into higher obesity classification levels, their health risks are also increasing. Obesity contributes to a multitude of serious diseases, functional impairments, and a higher risk of mortality (Allison et al., 2008). Comorbidities of obesity include, but

are not limited to, diabetes, metabolic syndrome, cardiovascular disease, cancer, foot pain and osteoarthritis.

Many developed countries are experiencing steady inclines in the proportion of obese adults. From 1991 to 2000 the prevalence of individuals who were overweight rose from 45% to 56.4% in the United States (Mokdad et al., 2001). Approximately thirty-eight million United States adults surveyed in 2000 were obese (Mokdad et al., 2001). The increase in obesity rates has been accompanied by an increase in the number of individuals with a BMI greater than 40 kg/m^2, growing from .9% in 1991 to 2.1% in 2000 (Mokdad et al., 2001). In 1991, only 4 states reported obesity rates greater than 15% with no states reporting a rate greater than 20% (Mokdad et al., 2001). However, by 2000, 22 states reported obesity rates of at least 20% (Mokdad et al., 2001). As obesity rates continue to rise in the United States and developed countries, disease prevalence and the cost of treatment can be expected to rise as well.

Major health concerns are surfacing for developed countries as the prevalence of obesity continues to rise. Obesity is linked to a wide array of mortality factors that are contributing to these health concerns. Hypertension is the most common syndrome related to weight and obesity in men and women (Must et al., 1999). An increase in the prevalence of Type 2 Diabetes Mellitus, gallbladder disease, and osteoarthritis for both genders has also been reported. An elevated prevalence of coronary heart disease exists in men for class I obesity and in women for all three classes of obesity (Must et al., 1999). For nonsmokers with no disease history, mortality rates are the highest among the heaviest men and women. This includes a 40-80% higher risk of dying from cancer (Calle et al., 1999).

Hypertension is the condition most commonly associated with heart disease. The Women's Health Study demonstrated that as blood pressure levels increased, women displayed a higher rate of incidence of Type II diabetes (Conen et al., 2007).Obesity, however, can be considered as an independent risk factor for CVD. Cardiovascular disease includes coronary heart disease, myocardial infarction, angina pectoris, congestive heart failure, stroke, hypertension, and atrial fibrillation (Must, 1999; Klein, 2004). Sudden causes of death related to CVD in Class III obese patients include dilated cardiomyopathy, severe coronary atherosclerosis, cardiomegaly, hypoplastic coronary arteries, and pulmonary emboli (Duflou et al., 1995).

Body weight can be used as a reliable predictor of the development of diabetes. Obesity is associated with an increase in inflammatory markers which is associated with insulin resistance (Klein et al., 2004). Increasing BMI levels had a significant effect on an increased risk of developing Type II diabetes (Conen et al., 2007). Type II diabetes is the condition in which the body does not produce adequate insulin or the insulin is not recognized by the cells (American Diabetes Association, 2009). Diabetes can lead to many life altering consequences such as deteriorating vision, nerve damage, and skin and foot sores that may ultimately lead to amputation.

Type 2 diabetes and hypertension are closely linked diseases and are both aspects of metabolic syndrome (Mokdad et al., 1998). Increased waist circumference, and dyslipidemia are also characteristics of metabolic syndrome. Individuals who are afflicted with metabolic syndrome are at a 1.5-3.0 times higher risk for suffering from coronary heart disease and stroke (Klein et al., 2004).

Furthermore, obesity is associated with pulmonary diseases such as asthma, chronic obstructive pulmonary disease, sleep disorders, hypoventilation syndrome, pulmonary thromboembolism, aspiration pneumonia and possible respiratory failure (McClean et al., 2008). Resta, et al. found that 51.5% of the obese population in their study had moderate to severe obstructive sleep apnea (Resta et al., 2001). The presence of this disorder often leads to loud snoring, restless sleep, daytime sleepiness, and headaches. These symptoms can lead to occupational and social harms (McClean et al., 2008). In addition to sleep disorders, obesity increases the severity of asthma (Shore, 2007). Moderate to severe asthma can greatly inhibit an individual's ability to perform physical activity and activities of daily living, potentially leading to an increase in obesity level and consequently reducing the quality of life.

In addition to diabetes, hypertension, cardiovascular disease, and pulmonary disease, BMI is strongly related to the development of cancer. Reeves, et al. found that among women, there is a positive relationship between increasing BMI and risk for the following cancers: all cancers, adenocarcinoma of the esophagus, squamous cell carcinoma of the esophagus, pancreatic cancer, lung cancer postmenopausal breast cancer, endometrial cancer, kidney cancer, and leukemia (Reeves et al., 2007). Individuals with a BMI over 40 kg/m^2 have a 52 and 62 percent higher death rate from all cancers for men and women, respectively, when compared to normal weight adults. It is estimated that 14 percent of deaths caused by cancer in men and 20 percent of deaths in women caused by cancer are related to being overweight and obesity (Calle et al., 2003).

Obesity can contribute to the development of mental disorders as well and is frequently related to major depressive disorder. Strine, et al. found that depression

severity is highly related to smoking, obesity, and physical inactivity. In addition, adults with a previous history of depression were more likely to smoke, to be obese, to be physically inactive, to binge drink and to drink heavily (Strine et al., 2008). While there has been no causal relationship discovered between obesity and depression, research shows that currently depressed individuals or those with lifetime diagnosis are 60% more likely to be obese and those with a lifetime diagnosis of anxiety are 30% more likely to be obese (Strine et al., 2008). Exercising and being physically active helps relieve symptoms of depression as well as serving to maintain current weight status or reduce obesity. While it seems that physical activity is often considered to cause obesity, it is possible that physical inactivity is a result of obesity and its limitations.

Several musculoskeletal disorders which may inhibit physical activity and activities of daily living while concurrently reducing the quality of life are associated with obesity. Increased load carriage that results from elevated adiposity generates increased ground reaction forces as well as increased sagittal plane torque at the hip, knee, and ankle. This increase in ground reaction forces and sagittal plane torques may play a role in the higher prevalence of musculoskeletal disorders among the obese population. While there is much conflicting research, links between obesity and foot pain in the plantar fascia and the plantar fat pad may exist (Prichasuk et al., 1994). Excessive load carriage may be the mechanism responsible for causing plantar fascia pain by producing more stress on the medial longitudinal arch (Prichasuk et al. 1994). In normal individuals, the fat pad is highly elastic and compressible, lending to its function of shock absorption and cushion during gait (Prichasuk et al., 1994). Overweight individuals with foot pain showed a higher level of heel-pad thickness and less heel-pad elasticity which

may, inherently, lead to a decreased level of shock absorption and cushion (Prichasuk et al., 1994).

A more definitive link has been identified between obesity and knee osteoarthritis. Knee osteoarthritis is exacerbated by the additional load carriage that results from obesity. Body mass index greater than 25 kg/m^2 is linked to having a more constant frontal plane knee adduction moment during the stance phase of gait (Harding et al., 2012). Obese individuals, classified as having a BMI greater than 30 kg/m^2, have lower sagittal plane knee extension moment magnitudes and smaller transverse plane knee rotation moments than their normal weight peers (Harding et al., 2012). Osteoarthritis results in a lower mean sagittal plane knee extension value for both the affected and unaffected legs. The affected leg, however, generally has a lower sagittal plane range of motion than the unaffected leg (Messier et al., 1992). Osteoarthritis also leads to a reduction in sagittal plane knee flexion in affected individuals compared to healthy individuals (Messier et al., 1992). Osteoarthritis patients have lower mean sagittal plane knee velocity than individuals free from osteoarthritis (Messier et al., 1992). Greater loading rates of the knee are found in individuals afflicted with osteoarthritis as well (Messier et al., 1992). Obesity appears to induce changes in gait mechanics which may ultimately result in the development of knee osteoarthritis.

Gait Characteristics of Normal, Lean Adults and Obese Adults

While gait appears to be a simple task for the healthy individual, it is an aspect of human movement defined by a myriad of characteristics. Defining components include stride length, stride frequency, stance phase, swing phase, and double support phase. Common measurements also include joint biomechanics such as joint angles, ground reaction forces, and torque and power created at the joints of the hip, knee, and ankle.

The movement of the lower limbs during the swing phase requires various muscle moments across the hip, knee, and ankle joints. During the first half of swing phase, inertial load is positive and the shank of the leg is in extension (Winter et al., 1978). Approximately 80% of the moment required to propel the shank is the result of forward deceleration of the knee joint and gravitational forces (Winter et al., 1978). The forward deceleration of the knee joint is a result of extensor deceleration at the hip (Winter et al, 1978). Inertial loads are reversed during the second half of swing phase as the shank decelerates (Winter et al., 1978). Eighty percent of this deceleration is caused by the muscle moments (Winter et al., 1978). Plantarflexors are the predominate muscle group activated during push-off (Winter et al., 1978). Hip flexors are the primary energy producers during the early portion of the swing phase (Winter et al., 1978). Deceleration of the leg is caused by absorption of energy by the hip extensors and knee flexors (Winter et al., 1978). A passive flow of energy from the foot and shank to the thigh and trunk is responsible for the final deceleration of the leg (Winter et al., 1978). During weight acceptance, stored energy from the trunk is transferred across the hip and knee and then absorbed by knee extensors and ankle dorsiflexors (Winter et al., 1978).

15

The function of the lower limb during stance phase is to resist collapse and to extend sufficiently to achieve the required push-off (Winter, 1980). Flexion at all three joints of the lower limb would result in collapse during the stance phase; therefore support of the body requires net extensor activity at these joints (Winter, 1980). Figure one demonstrates the muscle moments of the lower limb joints.

$$M_s = M_k - M_a - M_h$$
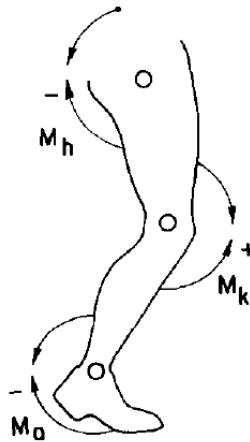
Fig. 1. Moments of force at the ankle, knee, and hip during stance, with extensor moments shown. Convention has counter-clockwise moment acting at proximal end of any segment as being positive. *Ms* is the total extensor moment acting on the lower limb segments. (Winter, 1980)

When one joint opposes or does not contribute to lower limb support one or both of the remaining joints will compensate for the non-contributing joints (Winter, 1980.) This theme is demonstrated by the variability between a young female walking at a natural

16

cadence (Figure 2) and an average of net supporting moment, *Ms*, of 12 normal subjects

(Figure 3).
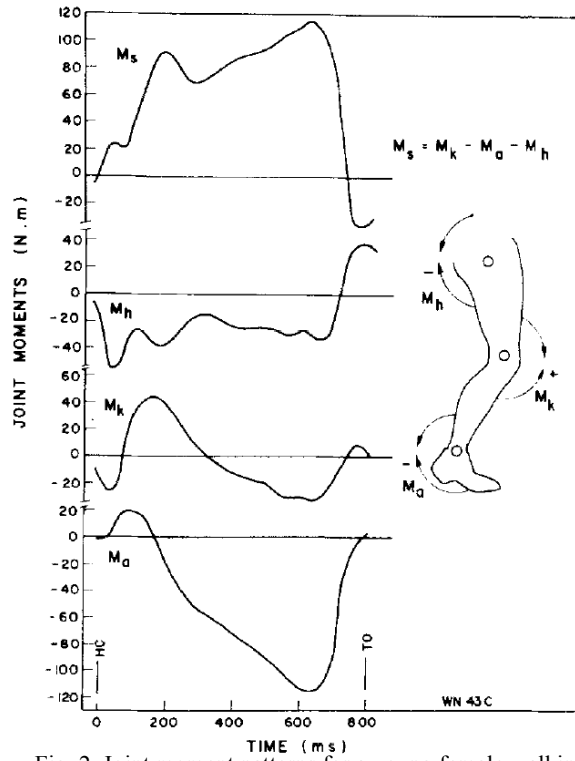
Fig. 2. Joint moment patterns for a young female walking at
natural cadence (Winter, 1980)
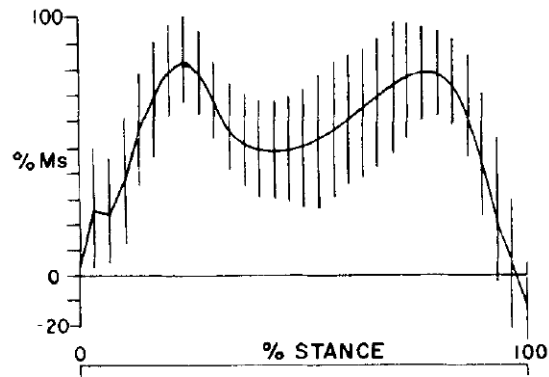
Fig. 3. Support moment, *Ms*, for 12 normals walking at comfortable cadence. Basic pattern shows a peak of support during weight acceptance followed by a second peak during late push-off (Winter, 1980)

Support during stance is achieved by a net extensor pattern. Therefore, kinetic assessments of gait should examine the total limb (Winter, 1980).

Massive weight loss, such as that produced by weight loss surgery, results in gait adjustments of obese individuals. Patients who lost 33% of their body mass have shown increases in swing time, longer strides, and faster walking speeds. These patients also showed an increase in hip range of motion in the swing phase as well as maximal knee flexion in early stance and a 40.5% shift of ankle joint function toward plantarflexion at a self-selected speed (Hortobagyi et al., 2011). Reductions in the first peak of vertical ground reaction force decreased 30.0% after massive weight loss (Hortobagyi et al., 2011). Massive weight loss appears to produce simple mass-related adaptations in gait mechanics as well as "mechanical plasticity in the form of a reorganization of lower extremity joint torques (Hortobagyi et al., 2011)."

Obese individuals have approximately 60% greater ground reaction forces than their normal weight counterparts at all speeds (Browning et al., 2007). Vertical and anteroposterior GRF both increase in proportion to body mass. Mediolateral GRF

18

displays a much larger increase than would be predicted based on body mass (Browning et al., 2007). This disproportion is potentially attributable to the larger step width seen in the obese individuals. The second medial peak formed when body mass is shifted to the contralateral stance leg does not decrease at a slower speed (Browning et al., 2007).

Diminutions in speed may be an adequate method to prevent musculoskeletal injury in obese individuals while walking. Reduced speeds result in smaller magnitudes of the first peak of vertical GRF despite having a smaller effect on the second peak vertical GRF. For example, vertical GRF peaks were reduced 15 and 6% when walking at 1.0 ms^{-1} versus 1.5 ms^{-1} (Browning et al., 2007). For both obese and normal weight individuals, peak anteroposterior GRF is approximately 40% smaller at 1.0 ms^{-1} than 1.5 ms^{-1}. The first peak of mediolateral GRF in obese individuals decreased by 23% when velocity was reduced from 1.5 ms^{-1} to 1.0 ms^{-1} (Browning et al., 2007). Figure 4 depicts the variations in GRF between obese and normal weight individuals at 1.5 ms^{-1} and obese individuals at six different speeds.
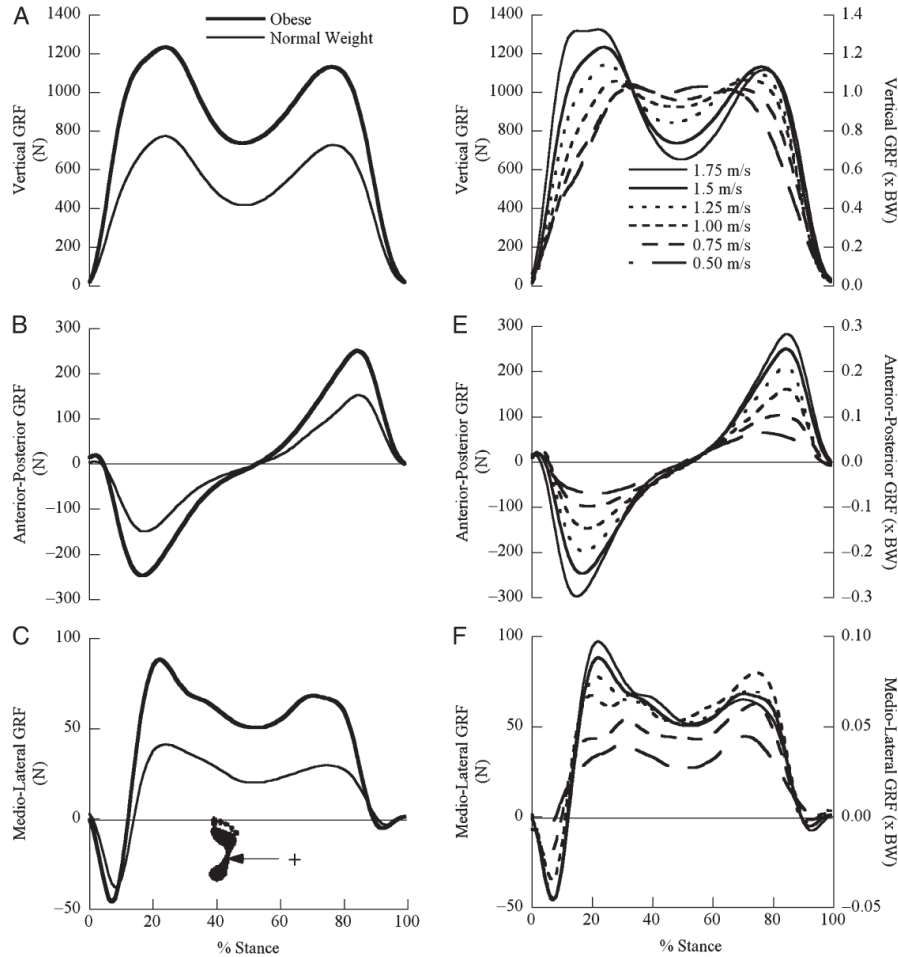
Fig.4. Mean vertical (A), anteroposterior (B), and mediolateral (C) ground-reaction forces (GRF) while walking at 1.5 ms^{-1} for the obese vs. normal-weight subjects, and vertical (D), anteroposterior (E), and mediolateral (F) GRF for the obese subjects walking at each speed. The cartoon in panel C shows the orientation of the mediolateral GRF relative to the foot. Compared with normal-weight subjects, the obese subjects had greater GRF, but in both groups GRF were smaller at the slower walking speeds (Browning et al., 2007).

Absolute net muscle moments (Figure 5) are also greater in obese individuals than normal weight individuals and are reduced with slower walking velocities (Browning, 2007). Peak hip extensor moment was significantly greater in obese individuals than normal weight individuals at all speeds except 0.75 ms^{-1} and peak extensor knee moments were significantly greater at 1.75 ms^{-1} (Browning et al., 2007). Peak knee extension moment was reduced by 43% when obese individuals walked at 1.0 ms^{-1} rather than 1.5 ms^{-1} (Browning et al., 2007).

20
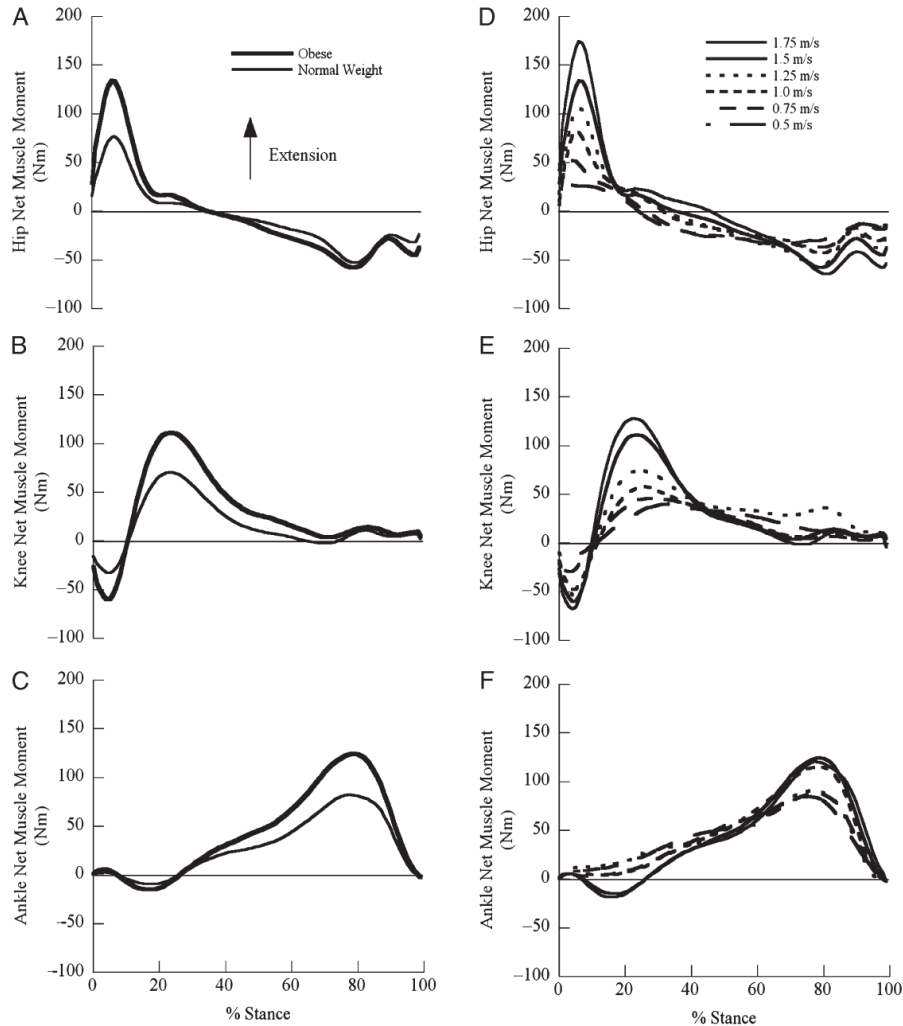
Fig. 5. Mean hip (A), knee (B), and ankle (C) net muscle moments during stance while walking at 1.5 ms^{-1} for the obese vs. normal-weight subjects, and hip (D), knee (E), and ankle (F) net muscle moments during stance for the obese subjects walking at each speed. Positive moments are extensor. Stance begins at right heel strike. Compared with normal-weight subjects, the obese subjects had greater net muscle moments, but in both groups net muscle moments were smaller at the slower walking speeds (Browning et al., 2007).

At a standard speed of 1.5 ms^{-1}, normal weight individuals walk with a stride length of approximately 1.49 m and an approximate frequency of 1.01 Hz. For the same individuals, 61.5% of the gait cycle was spent in the stance phase, 38.5% of the cycle was spent in swing phase, and 21.8% of the cycle was spent in double support phase. Stride width was found to be about 30% greater in obese subjects than lean subjects (Browning et al., 2007). Additional research shows similar findings of stride frequency, stance phase, and swing phase (DeVita et al., 2003). Table 1 demonstrates these findings.

	Stride Length (m)		Stride Frequency (Hz)		Stance (% Cycle)		Swing (% Cycle)	
	O	NW	O	NW	O	NW	O	NW
Browning, et al.	1.51	1.49	1.00	1.01	64.10	61.50	35.90	38.50
DeVita, et al.	1.44	1.46	1.05	1.05	63.4	61.6	36.6	38.4

Table 1. Stride Characteristics of Obese and Normal Weight Individuals at a Standard Speed of 1.5 ms^{-1}

While Browning and Kram found no difference between lean and obese individuals for stride length and stride frequency, their results did demonstrate that obese individuals spent longer times in stance phase and double support phase and less time in swing phase. Stride length and frequency for both groups increased as speed increased. Stance phase time and double support phase time are inversely related to speed while swing phase time and gait speed are positively related (Browning et al., 2007). Separate research has shown that obese individuals walk with a 7% shorter step length and 11% lower step rate while walking at a 16% slower self-selected velocity than normal weight individuals (DeVita et al., 2003). At this velocity, obese individuals had a 3% shorter relative swing phase and a 2% longer relative stance phase (DeVita et al., 2003).

Precise, biomechanical differences of gait between normal weight individuals and moderately obese individuals have not been definitively defined. However, adaptations have been noted as individuals move from a level of moderate obesity into Class III obesity. Normal weight adults and moderately obese adults ambulate with no significant differences in joint angles of the hip, knee, and ankle (Browning et al., 2007). At slower speeds, both normal weight and moderately obese adults extend more at the hip and knee are generally more extended, whereas the ankle is more plantarflexed during stance and more dorsiflexed during swing (Browning et al., 2007). On the contrary, there is a

difference in angular joint position between the normal weight and Class III obese individuals. Class III obese individuals increase knee flexion in early stance by 12% and ankle plantarflexion at toe off by 11% to increase velocity (DeVita et al., 2003). Obese individuals were found to be more erect, demonstrating approximately 5º more hip extension throughout stance phase, approximately 8º less knee flexion in early stance, and approximately 4º less knee flexion throughout the entire stance phase (DeVita et al., 2003). These individuals were approximately 6º more plantarflexed throughout stance phase and approximately 7º more plantarflexed at toe off than normal weight individuals (DeVita et al., 2003). Obese individuals increased their extensor angular impulse at the knee by 17% and plantarflexor angular impulse at the ankle by 12% (DeVita et al., 2003). At a standard speed of 1.5 ms^{-1}, these larger impulses produced 68% more negative work at the knee in early stance and 11% more positive work at the ankle throughout stance while hip torque and power were not different than that of normal weight individuals (DeVita et al., 2003). Subjects used 89% more plantarflexor angular impulse that produced 61% more work at the ankle joint (DeVita et al., 2003).
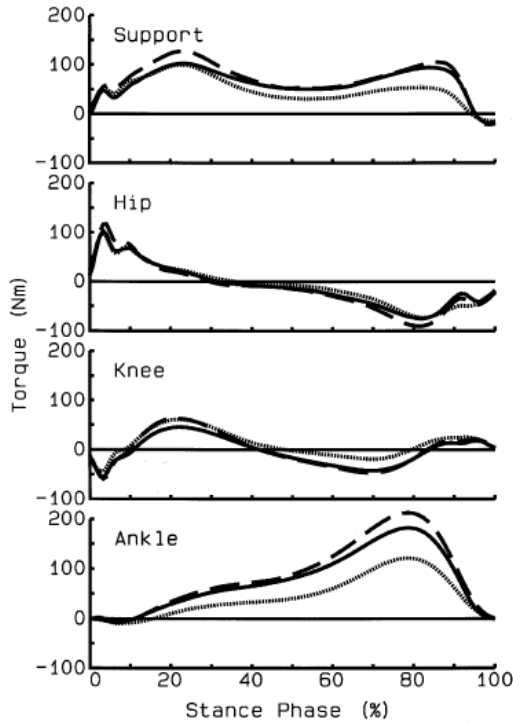
Fig. 6. Sagittal plane joint torques during the stance phase for obese (self-selected speed: solid line; standard speed: dashed line) and lean (dotted line) participants. Positive values are extensor or plantarflexor and negative values are flexor or dorsiflexor torques. Obese and lean had equivalent knee extensor angular impulses in early stance (area under curve from B10% to B40% of stance) at the standard speed and obese had less angular impulse at the slower, self-selected speed. Obese had larger ankle plantarflexor torque compared to lean at the same speed and larger torque in the faster, standard speed compared to the self-selected speed. Hip torques were equivalent in all conditions. Support toque was larger in obese compared to lean and larger in early stance at the faster compared to slower speeds in obese (DeVita et al., 2003).
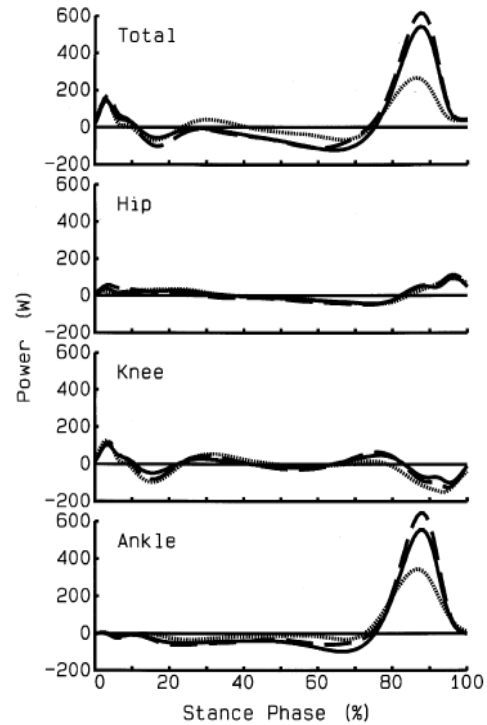
Fig. 7.Sagittal plane joint power during the stance phase for obese (self-selected speed: solid line; standard speed: dashed line) and lean (dotted line) participants. Positive values indicate energy generation through concentric contractions and negative values indicate energy absorption through eccentric contractions. Obese and lean had equivalent power at the hip and knee joints at the same walking speed but less knee power at the slower speed. Obese had larger ankle power at the faster compared to slower speeds and compared to lean at the same speed. Total power was larger in obese compared to lean, due mostly to differences in ankle power (DeVita et al., 2003).

Obese individuals walked with greater torques than normal weight individuals. This increase in torque was a byproduct of greater GRF while walking with similar joint kinematics (Browning et al., 2007). While obese individuals walked with similar stride characteristics to normal weight individuals, they displayed a linear increase in ground reaction forces, joint torques, and BMI (Browning et al., 2007). There is a qualitative,

behavioral change in gait, potentially mass related, that occurs in Class III obese

individuals at an unidentified cut-off point as defined by BMI (DeVita et al., 2003). This

change results in a gait pattern which elicits a more erect gait with torque and power at

the hip and knee being nearly identical to that of normal weight individuals yet larger

plantarflexor torque and power at the ankle (DeVita et al., 2003).

Measurement Techniques in Gait Analysis

Gait is measured using motion capture systems and force plates. While there are

many ways to collect three-dimensional kinematics, modern three-dimensional motion

capture utilizes infrared markers and a camera tracking system to record the movement of

an individual's joints and segments in space. Three-dimensional systems allow wide

array of research including the computation of joint angles and the location of joint

centers. Force plates are used to measure many different types of forces. This thesis will

examine the use of a force plate to measure vertical ground reaction forces during gait.

While three-dimensional analysis can give us 3D coordinates of body segments in

space, the simultaneous use of force plates allows the acquisition of force vectors and

ground reaction forces during movement as well. In the 1930's, Herbert Elftman was able

to utilize a force plate to examine the muscle function in a human leg while walking

(Elftman, 1948). Elftman was able to show that the application of ground reaction forces

is initially placed on the foot in an upward and backward pattern followed by an upward

and forward pattern (Elftman, 1948). The use of ground reaction forces and the point of

its application combined with video images of the displacement of the body over time

allow the study of kinetics of the leg without the necessity of considering the entire body.

Elftman also stated that his methods of determining the instantaneous value of the forces and torques acting on each part of the leg allows the calculation of the sum of the torques exerted by the muscles on the foot, shank, and thigh (Elftman, 1948). Joint torques are frequently estimated using the process of inverse dynamics. Forces inside our joints and torques produced by muscles are calculated using ground reaction forces and body positions coupled with anthropometric characteristics. In the 1930's, Elftman also used inverse dynamics to calculate the torque of one-joint muscles and, although they are not actually present in the lower limb, create a model of three-joint muscles (Elftman, 1939).

Free body diagrams are created to represent body segments under investigation. These models allow the association of positional and inertial properties (Cappozzo et al., 1984). Positional properties convey the information needed to locate a segment at any point in time in reference to an arbitrarily chosen observer. An observer is defined by a local coordinate system rigid with the segment and a global coordinate system rigid with the observer (Cappozzo et al., 1984). Stereometric techniques allow the determination of the instantaneous position of points fixed in a moving segment and, therefore, allow simple vector calculations of the segment transformation matrix and position vector (Cappozzo et al., 1984). These techniques require target points that can be represented by skin markers.

The majority of modern marker placement sets are based from the Helen Hayes model (Kadaba et al., 1990). This system was designed using minimal markers to simplify the identification of marker trajectories. The selected marker positions were chosen to satisfy rigid body assumptions (Kadaba et al., 1990). In the Helen Hayes marker system, two markers were placed on the right and left anterior superior iliac

spines, one marker was placed on a stick 10 cm long extending from the top of the

sacrum in the spinal plane, and one marker is also placed on the greater trochanter, lateral

malleolus, and the space between the second and third metatarsal heads (Kadaba et al.,

1990). An additional cuff was positioned on the midthigh and another on the midshank

with 7 cm long wands with markers attached at the tip (Kadaba et al., 1990).
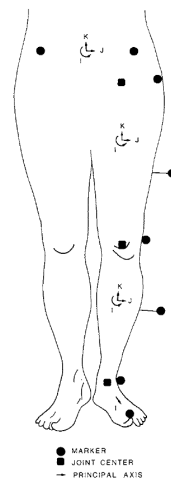
Fig. 8. Marker configuration and embedded
coordinate systems (Kadaba et al., 1990).

Lower limb joints work across all three planes during gait. The majority of motion

occurs in the sagittal plane; however there is a significant amount of movement within

the frontal plane as well. The hip experiences close to the same amount of torque in the

frontal plane as in the sagittal plane but much less in the transverse plane. Subjects

showed more frontal plane torque at the knee joint than the sagittal and transverse. The

ankle joint demonstrated a much larger torque in the sagittal plane than either the frontal

or the transverse (Eng et al., 1995). Figures 9-11 demonstrate these findings. These

27

results confirmed the necessity of testing gait in a three-dimensional model rather than a planar model (Eng et al., 1995).

Fig. 9. Hip joint moments normalized to body mass in the frontal, transverse, and sagittal planes. The dashed line represents one standard deviation above and below the mean.
(Eng et al., 1995)

28

Fig. 10. Knee joint moments normalized to body mass in the frontal, transverse, and sagittal planes. The dashed line represents one standard deviation above and below the mean.
(Eng et al., 1995)

29

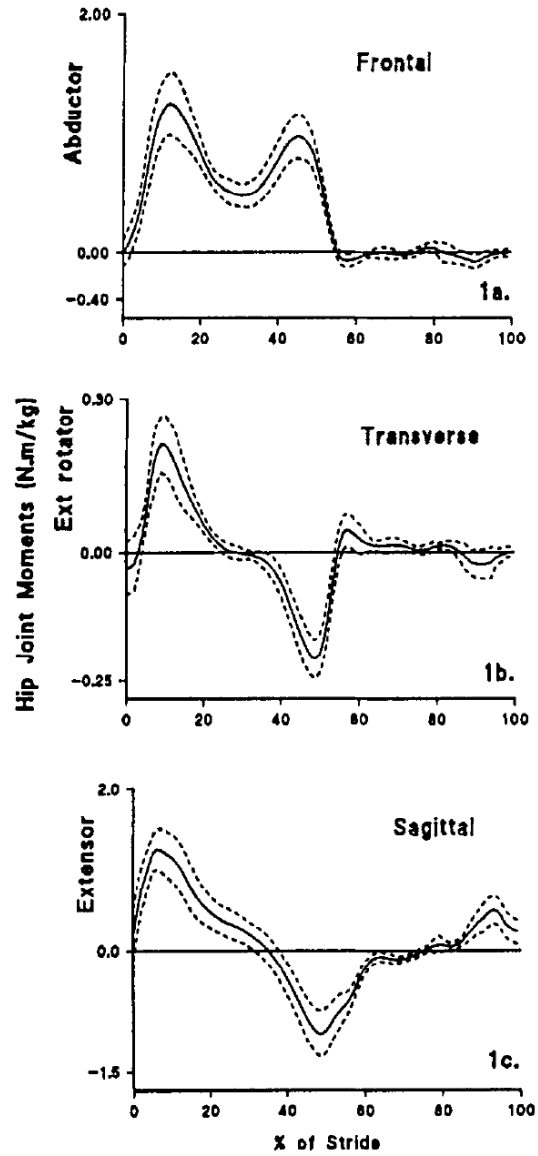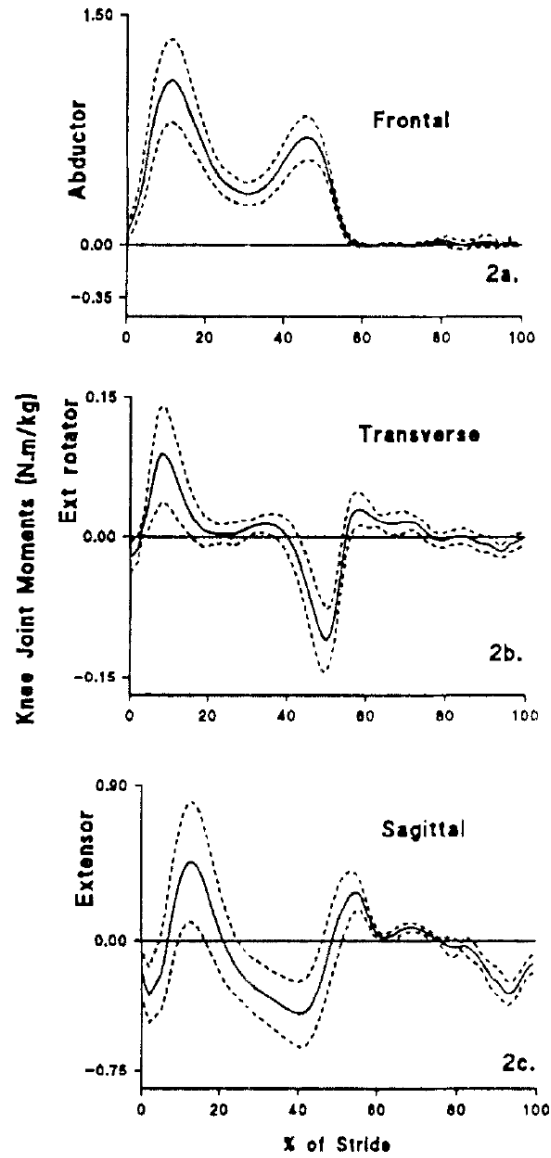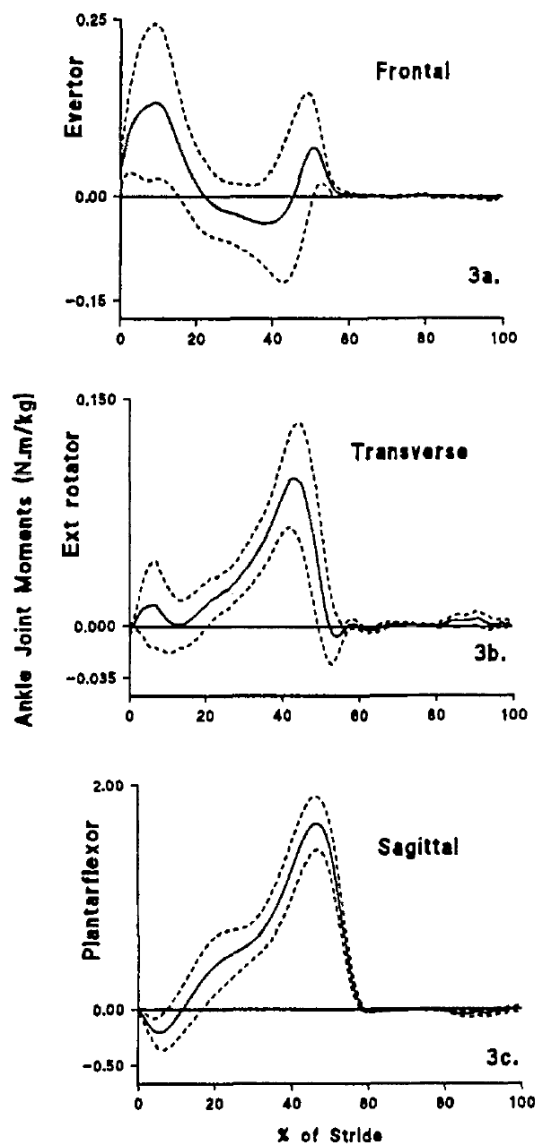**Fig. 11. Ankle joint moments normalized to body mass in the frontal, transverse, and sagittal planes. The dashed line represents one standard deviation above and below the mean.**
(Eng et al., 1995)

30

Summary

Obesity has become an epidemic in well-developed countries. It is accompanied by health altering diseases including, but not limited to, cancer, cardiovascular disease, diabetes, and osteoarthritis (Allison et al., 2008). Osteoarthritis alone may create alterations in gait patterns. Obesity, as well as being a potential risk factor for osteoarthritis, yields its own characteristics that result in gait alterations.

Gait is a complex movement pattern defined by such measurements as kinematics, joint biomechanics, and ground reaction forces. Obese individuals have been found to experience greater ground reaction forces and torques than normal weight individuals. Moderately obese individuals with a BMI of approximately 35 kg/m^2 walk with similar design as lean individuals while exerting higher ground reaction forces and joint torques (Browning et al., 2007). However, Class III obese individuals with a BMI closer to 45 kg/m^2, display similar joint angles at the hip and knee but more plantarflexion at the ankle than normal weight individuals. There appears to be a yet un-defined cut-off point, as defined by BMI, where obese individuals shift from a linear adaptation to body mass to a qualitative behavioral adaptation. Additional research is needed to determine what level of obesity triggers individuals to make a change in muscle activation to compensate for carrying a larger load.

Chapter 3: Methodology

Subject Characteristics

Twenty-one subjects were recruited for this study using newspaper ads and flyers located on the East Carolina University campus as well as in local businesses. The East Carolina University listserve was utilized to send e-mail announcements to faculty and staff. These e-mails initially aimed at recruiting individuals of all BMI levels. During our study these e-mails were resent periodically, targeting more specific BMI categories by specifying weight requirements such as weighing more than two hundred pounds. Flyers were also distributed to bariatric patients at Physicians East in Greenville, NC. Phone interviews were conducted to determine eligibility for the study. Inclusion criteria specified that subjects must be between the ages of 18 and 60 years old with a $20 \text{ kg/m}^2 \leq$ BMI $\leq 50 \text{ kg/m}^2$. Our goal was to recruit one subject per BMI unit for our target range. Despite our purposeful recruitment, we were unable to procure and subjects with a BMI above 44kg/m^2 and only a limited sample of subjects above a BMI of 37 kg/m^2. A BMI range of 18 kg/m^2 to 44 kg/m^2 characterized our final subject sample.

In order to participate in this study, subjects were required to be apparently healthy with the exception of high BMI. Individuals were excluded if they had any neurological, musculoskeletal, neuromuscular, or physiological disabilities. Previous surgeries to the lower limbs were also considered as exclusion criteria. Qualifying subjects were required to sign a written informed consent form (see Appendix) before participating. We tested subjects evenly distributed within the range of body mass indexes between 18 kg/m^2 and 44 kg/m^2. Table 2 represents subject characteristics.

Subject	Age	Mass (kg)	Height (m)	BMI (kg/m^2)
S01	47	81.7	1.69	28.6
S02	43	95.7	1.67	34.4
S03	42	98.8	1.76	32.0
S04	51	82.1	1.86	23.7
S05	43	86.4	1.68	30.6
S06	33	78.6	1.62	30.0
S08	44	68.4	1.60	26.6
S09	58	89.3	1.66	32.5
S10	47	65.8	1.55	27.3
S11	35	54.5	1.66	19.8
S12	51	103.4	1.68	36.8
S13	26	80.0	1.56	32.7
S14	46	69.0	1.61	26.5
S16	32	108.6	1.87	31.2
S17	29	45.0	1.57	18.3
S18	52	56.8	1.57	23.1
S19	45	93.5	1.61	36.3
S20	44	165.3	1.94	44.0
S21	59	94.4	1.72	31.9
S22	30	81.8	1.65	30.0
S23	25	74.8	1.81	22.7
Mean	41.5	84.5	1.70	29.5
SD	10.8	24.7	0.10	6.1

Table 2: Subject Characteristics

Equipment and Instruments

A phone interview was used to determine eligibility. Subjects also completed the

SF-36. This 8-scale, 36 question short-form health survey was used to assess physical and

mental health (Ware et al., 1992). A copy of this form can be located in the Appendix. A

timed 20m walking trial was conducted in a hallway to evaluate functional capabilities.

Quadriceps strength was evaluated with isokinetic strength testing using a

HUMAC Norm, isokinetic dynamometer (Computer Sports Medicine, Inc, Ma) at

60° sec^{-1} and 150° sec^{-1}. Strength testing was performed to investigate whether the

33

hypothesized non-linear relationship between BMI and gait mechanics was mirrored by such a relationship between BMI and strength.

Biomechanical gait analyses were performed using a 15 m level walkway instrumented with an AMTI force plate used to collect ground reaction forces (AMTI Model LG-6, Newton, MA). Ground reaction forces were measured at 960 Hz and a gain of 4000. Gait kinematic and kinetic data were recorded with the use of eight Qualisys ProReflex MCU 240 cameras (Qualisys Medical AB, Gothenburg, Sweden) at 120 Hz. An infrared timing system was used to control for the walking speed of each subject (Brower timing system, model IRD-T175, Salt Lake City, Utah). Data were collected using Qualisys Track Manager Software (Innovision Systems Inc., Columbiaville, MI) and analyzed using Visual 3D (C-Motion Inc., Rockwille, MD). Height and mass for all subjects were measured using a Seca 703 scale (Seca gmbn & Co.kg, Hamburg, Germany).

Procedures

Participants were tested on two occasions for approximately 45 minutes on day 1 and approximately an hour and 20 minutes on day 2. Testing was conducted in the Biomechanics Laboratory, room 332 of Ward Sports Medicine Building, East Carolina University, Greenville, NC. Before participation, subjects were required to read and sign an informed consent form. All data were collected while subjects were wearing tight fitting athletic style shorts and tennis shoes.

On day 1, subjects completed a functional test. This was comprised of a timed 20m walk in a hallway. Upon successful completion, subjects were isokinetically tested

for quadriceps strength using the HUMAC isokinetic dynamometer. Velcro straps were used to stabilize the torso and thigh. Subjects were given a warm-up period to habituate themselves with the equipment and protocol. The input axis of the dynamometer was aligned with the knee joint. The right leg was tested for each subject with five trials at an angular velocity of $60^{\circ}sec^{-1}$ and five trials at an angular velocity of $150^{\circ}sec^{-1}$. The maximum peak torques for extension and flexion at each angular velocity were used for analysis.

Subjects returned to the lab approximately three days after strength testing. DXA scans were performed in the East Carolina University F.I.T.T. Laboratory. Upon return to the Biomechanics Laboratory, reflective markers were placed on the subject using a Helen Hayes set-up. Fifteen individual markers and three marker plates were situated on the participants in the following manner: individual markers were placed on the right foot on the first and fifth metatarsal heads, heel, and medial and lateral malleoli, a three marker plate was used on the superior aspect of the right foot, a four marker plate was attached to the shank, individual markers were placed on the medial and lateral femoral condyles, a four marker plate was attached to the thigh, individual markers were used to identify the right and left greater trochanters, right and left ASIS, right and left PSIS, and right and left iliac crests for a 5 second static calibration. During this calibration trial, subjects stood stationary in the center of the collection field with feet shoulder width apart and arms crossed over the chest. After the static calibration, the markers of the metatarsals, malleoli, knee, greater trochanters, and iliac crests were removed. This modified marker-setup was used for a static calibration trial as well as all movement trials.

Subjects were allowed to practice walking along the walkway to obtain the correct speed. These practice trials also served as a tool for subjects to become comfortable in the environment and to reduce anxiety. Subjects were never instructed to walk at a specific speed. Instructions for the self-selected, normal pace included being instructed to walk at a pace the subject would use when walking to work or an appointment. This created an "intentional" self-selected speed during which the subjects walked as if they were heading some place. When walking at a safe-maximum speed, subjects were instructed to walk as fast as possible without feeling as though they were going to run or fall. Five trials were collected at a self-selected speed and five trials were collected at a safe-maximum walking speed. Subjects were never instructed to strike the force plate while walking. Trials in which the participant's right foot did not fully contact the force plate or the left foot contacted the plate were discarded. Any trials in which participants made noticeable changes to their gait were also discarded. Participants were given the opportunity to rest between trials if needed.

Dual X-Ray Absorptiometry

Dual x-ray absorptiometry (DXA) was used to obtain subject specific body segment inertial parameters (BSIPs). We conjectured that as obesity increases, variations in body segment mass, as a percentage of total body mass, would produce meaningful differences in BSIPs and, in turn, alter lower extremity joint torques. The largest torque variations were expected at the hip joint and knee joint. We failed to identify a significant correlation however between BMI and percent body mass of the pelvis, right thigh, or right shank. A significant correlation was identified between BMI and the right foot. Additionally, we were unable to find a significant correlation between total body mass

and percent body mass of the right thigh or right shank. A significant correlation was found between total body mass and percent body mass of the pelvis and the right foot. The DXA procedure altered the thigh relative weighting by +4.09% but had relatively small differences for the pelvis, shank, and foot. The relationships between segment mass and BMI were low and nonsignificnat overall. However in more obese individuals, a difference was found between the segment masses found using the DXA and the standard segment measure typically utilized. Therefore, we applied this DXA method to all subjects. These data are presented in Table 4.

	Mean (% Total Body Mass)	Standard Deviation	Correlation with BMI	Correlation with Total Body Mass
Pelvis	14.45	2.79	0.361	0.539*
Right Thigh	14.09	1.22	0.144	0.047
Right Shank	4.29	0.65	-0.047	0.152
Right Foot	0.98	0.18	-0.618*	-0.487*

Table 3. Body Segment Parameters Correlation with BMI and Total Body Mass, * $p < 0.05$

Data Reduction

Data from each subject were processed using Qualisys Track Manager Software which provided data in the global coordinate system (GCS) for all subjects and trials. Position data were obtained for each subject with respect to the coordinate system of the Biomechanics Laboratory. QTM identified markers by anatomical location and correct trajectories after gaps for missing data have been filled. Files were exported to Visual 3D after filling gaps for marker fallout. A linked rigid-segment system was used to build a

model of the lower extremity. A static standing trial, known as a calibration trial, was performed for each subject. This trial was used to create an individualized model per subject enabling the location of joint centers, location of the segment center of mass, definition of the local coordinate system of each segment (LCS), and location of the reflective markers in the coordinate system. Joint centers were located at fifty percent of the distance between the medial and lateral calibration markers for knee and ankle joints. The hip joint was located twenty five percent of the distance between the right and left greater trochanters.

A pipeline was used for signal processing. A low pass digital filter was used with cutoff frequencies of marker positions (6 Hz), analogs (45 Hz), ground reaction forces (45Hz), center of pressure (45 Hz), and free moments (45 Hz).

The analysis then produced for each trial: stride length, walking velocity and joint angular positions and velocities of the hip, knee and ankle. Joint positions, gait velocity, stride length, and stride velocity of the segments or joints are referred to as the kinematics of gait. Angular positions and velocities of the hip, knee, and ankle joints were calculated using a series of commands in Visual 3D. Joint torques were processed with a second low pass filter at 10 Hz. This second filter removed noise that is seen at initial heel strike. Joint angular displacements were calculated to find joint position data. Calculation of the angle between two adjacent segments allowed for the calculation of angular displacement. Joint angular displacement (θ) was calculated using the difference of the initial (θ_i) and final (θ_f) joint angular positions.

$$(\theta) = \theta_f - \theta_i \qquad\qquad (1)$$

Velocities were calculated using joint angular displacement per unit of time.

38

$$\Omega_i = (\theta_{i+1} + \theta_{i-1})/(2*\Delta t) \qquad (2)$$

Joint torques and powers were then calculated at the ankle, knee, and hip using the following procedures. Joint torque was calculated by first obtaining joint reaction forces (JRF) at the ankle using the following equation where segment mass is m, acceleration of the segment center of mass is a_{cm}, gravity force vector is mg, and ground reaction force is f_{grf}:

$$JRF_{ankle} = ma_{cm} - mg - f_{grf} \qquad (3)$$

Joint torques (JM) were then found using the following equation:

$$JM_{Ankle} = I\alpha - (d1 * JRF_{Ankle}) - (d2 * F_{GRF}) - t \qquad (4)$$

Where I is the moment of intertia, α is angular acceleration, d1 * JRF_{Ankle} describes the moment as a result of JRF, d2 * F_{GRF} is described from the GRF, and t is the ground reaction torque. This equation was then reversed with JRF replacing GRF, shifting ground reaction forces, torques, center of pressure, force on the segment due to gravity (mg), segment center of mass accelerations (ma_{cm}), proximal (JM_{Prox}) and distal (JM_{Distal}) moment arms, and proximal (JRF_{Prox}) and distal (JRF_{Distal}) joint center locations into the local coordinate system in order to solve for the knee and hip. These equations are represented as the following:

$$JRF_{Prox} = ma_{cm} - mg - JRF_{Distal} \qquad (5)$$

$$JM_{Prox} = I\alpha - (d1 * F_{JRF\_Prox}) - (d2 * F_{JRF\_Distal}) - JM_{Distal} \qquad (6)$$

Data Analysis

Scores on the SF-36 and 20 m functional walking test were correlated with BMI. A correlation was measured between mass and peak isokinetic knee torque using the data from the HUMAC. Relationships between peak vertical ground reaction forces, peak hip, knee, and ankle joint torques, and peak hip, knee and ankle joint powers with respect to BMI were analyzed using Pearson Product Moment Correlation and regression analyses. Linear regressions were fitted to vertical ground reaction force, peak hip, knee, and ankle joint torques, and peak hip, knee, and ankle joint powers. Significance was tested at the $p < .05$ level. Variables found to have a significant correlation with BMI were further evaluated by normalizing to mass. This additional analysis allowed the assessment of the relationship as being mass, or BMI, driven as opposed to being driven by an unknown variable. For our subject sample, the correlation of BMI and mass was 0.885, this strong correlation supports the idea of normalizing for mass.

The allowed variance of age and speed among participants could influence an identified relationships between joint torques and powers and ground reaction forces with BMI. Therefore, the relationships between BMI and our evaluated variables were explored after removing variation due to either walking speed or age.

Chapter 4: Results

The primary hypothesis of this study was that there is a critical magnitude of obesity, as measured with Body Mass Index, above which humans make behavioral adaptations to their walking biomechanics. It was speculated that these behavioral adaptations enable Class III obese individuals to modulate hip and knee torques and powers to values exhibited by lean individuals while simultaneously increasing torque and power at the ankle despite having excessive body mass and ground reaction forces. This chapter presents the gait biomechanics of healthy adults with BMI values between 20 kg/m^2 and 44kg/m^2 tested at a self-selected normal speed and a self-selected safe maximum speed. We originally proposed to test people with BMIs up to 50 kg/m^2, but we were unable to recruit individuals above 44 kg/m^2. The final sample size was 21 participants.

The purpose of this study was to identify the relationships between ground reaction forces, joint torques, and joint powers at the hip, knee, and ankle and BMI in a group of people with BMIs ranging from 18-44 kg/m^2, while walking at a self-selected normal speed and at a safe self-selected maximum speed. Dual x-ray absorptiometry was used to calculate the percent of body mass for the pelvis, right thigh, right shank, and right foot. Isokinetic concentric knee flexion and extension were tested using a HUMAC isokinetic dynamometer.

20 Meter Walk Test and SF-36

A timed 20 meter (m) walk test and the SF-36 were used to assess the physical competency of subjects. The mean and standard deviation for the timed 20m walk were 13.7s ± 1.8s. Significant correlations were found between walk time and BMI as well as

walk time and total body mass. The average physical competence score (PCS) and mental

competence scores (MCS) were 53.6 and 53.8, respectively, with standard deviations of

53.1 and 6.8. No significant correlations were found between PCS or MCS and BMI.

These data are presented in Table 3.

	Mean	Standard Deviation	Correlation with BMI	Correlation with Total Body Mass
20 m Walk	13.7 sec	1.8 sec	0.463*	0.649*
SF-36 PCS	53.6	5.1	-0.407	-0.268
SF-36 MCS	53.8	6.8	-0.031	-0.222

Table 4. 20m Walk Test and SF-36 Results and Correlation with BMI and Total Body Mass, *p < 0.05

Isokinetic Strength Testing

Right leg knee flexor and extensor strength were tested isokinetically at 60°/s and

150°/s using a HUMAC isokinetic dynamometer. These data were used to assess any

changes in strength associated with BMI and total body mass. We failed to identify a

significant correlation for any condition at the $p < 0.05$ level. These data are presented in
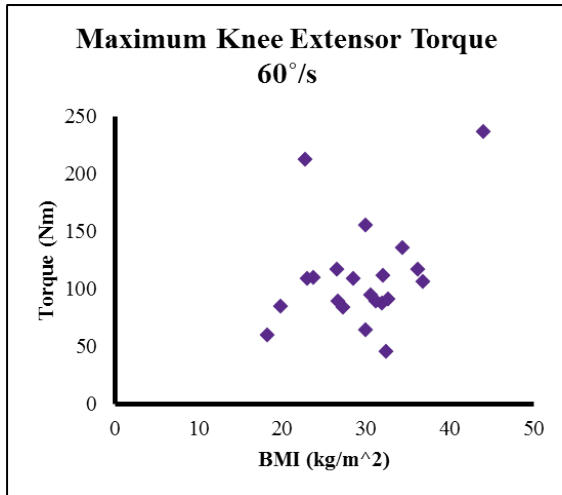
Figures 12 through 15.

Figure 12. Non-significant relationship between maximum knee extensor torque at 60˚/s and BMI, $r = 0.336$, $p > 0.05$
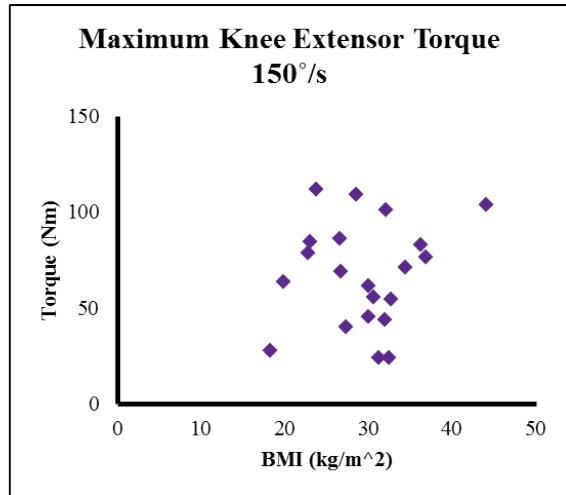

Figure 13. Non-significant relationship between maximum knee extensor torque at 150˚/s and BMI, $r = 0.144$, $p > 0.05$
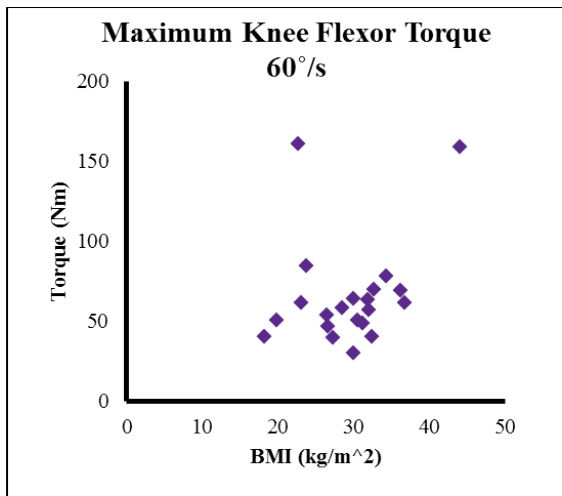

Figure 14. Non-significant relationship between maximum knee flexor torque at 60˚/s and BMI, $r = 0.269$, $p > 0.05$
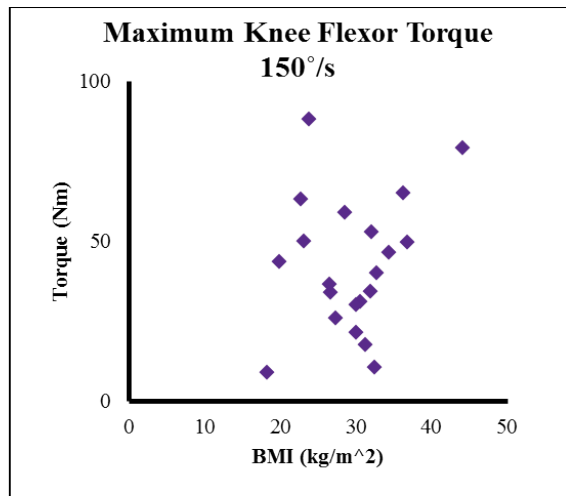

Figure 15. Non-significant relationship between maximum knee flexor torque at 150˚/s and BMI, $r = 0.224$, $p > 0.05$

43

Gait Biomechanics

The purpose of this study was to identify the relationships between ground reaction forces, joint torques, and joint powers at the hip, knee, and ankle and BMI while walking at a self-selected normal speed and at a safe self-selected maximum speed for healthy adults with BMI's ranging from 20-50 kg/m^2. Due to recruitment difficulties, only one subject with a BMI above 40 kg/m^2 was included in this data set.

Hip Joint

We expected to find a curvilinear relationship between hip joint torques and powers and BMI. More specifically, we expected to see an increase in torques and powers as BMI increased to some BMI at ~35 kg/m^2 after which point, torques and powers would decrease and begin to mimic those seen in normal weight individuals.

At the self-selected speed, maximum hip extensor torque was found to be significantly correlated with BMI. We failed to identify a significant correlation between maximum positive power at the hip joint and BMI while subjects ambulated at their self-selected normal speed. These data are depicted in Figures 16 and 17. The significant relationship between maximum hip extensor torque and BMI at the self-selected speed was further evaluated by normalizing to mass. This produced a non-significant correlation with $r = 0.292$. The lack of a significant relationship after normalizing for mass verifies the relationship is, indeed, mass driven opposed to being driven by unidentified variables.
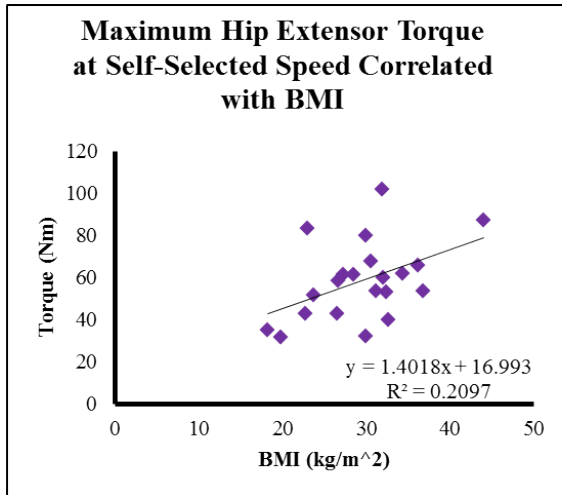
44

Maximum Hip Extensor Torque at Self-Selected Speed Correlated with BMI

$y = 1.4018x + 16.993$
$R^2 = 0.2097$

Figure 16. Significant relationship between maximum peak hip extensor torque and BMI, $r = 0.458$, $p < 0.05$

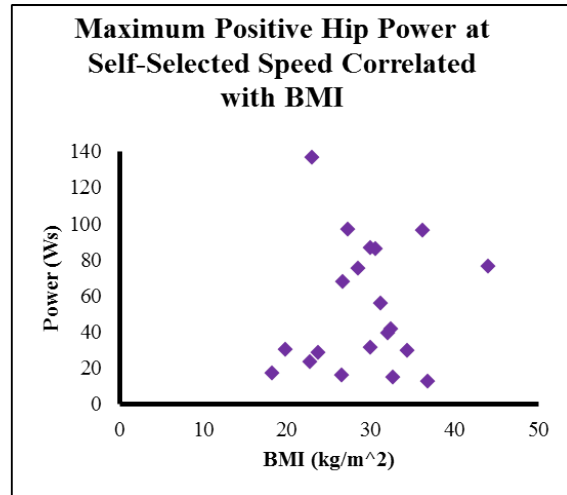**Maximum Positive Hip Power at Self-Selected Speed Correlated with BMI**

Figure 17. Non-significant relationship between maximum positive hip power and BMI, $r = 0.143$, $p > 0.05$

Similar results were found at the hip joint with subjects walking at a safe-maximum speed. A significant correlation was found between maximum hip extensor torque and BMI at a safe-maximum speed. We failed to find a significant relationship between maximum positive hip power and BMI at a safe-maximum speed. These data are shown in Figures 18 and 19. Additional analysis with mass normalization found a non-significant relationship between maximum hip extensor torque and BMI at the safe-maximum speed, indicating the relationship between hip extensor torque and BMI is mass driven.
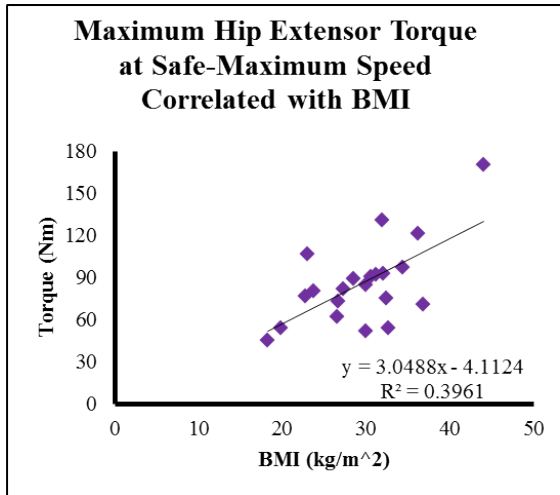
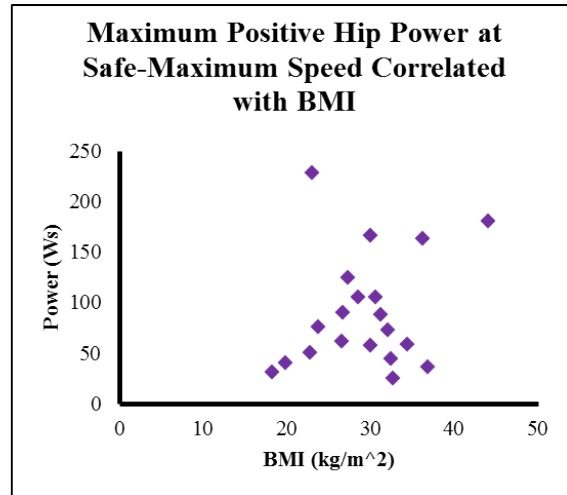Figure 18. Significant relationship between maximum peak hip extensor torque and BMI, r = .629, p < 0.05

Figure 19. Non-significant relationship between maximum positive hip power and BMI, r = 0.237, p > 0.05

Knee Joint

We expected to find a similar result at the knee joint as expected at the hip joint. It was hypothesized that knee joint torques and powers would increase to a BMI of approximately 35 kg/m^2 and then begin to decrease and mimic the torques and powers of normal weight individuals as BMI continued to increase. Our subject sample failed to present a significant relationship between knee joint extensor torque and maximum negative knee joint power at the self-selected speed. These data are presented in Figures 20 and 21.
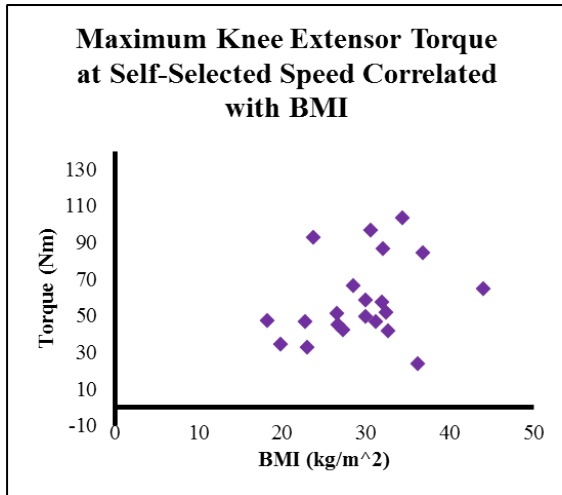
46

Figure 20. Non-significant relationship between maximum peak knee extensor torque and BMI, r = .284, p > 0.05

Figure 21. Non-significant relationship between maximum negative knee power and BMI, r = 0.188, p > 0.05

Additionally, we failed to find a significant relationship between maximum knee extensor torque and maximum negative knee power at the safe-maximum speed, seen in figures 22 and 23.

Figure 22. Non-significant relationship between maximum peak knee extensor torque and BMI, r = .251, p > 0.05

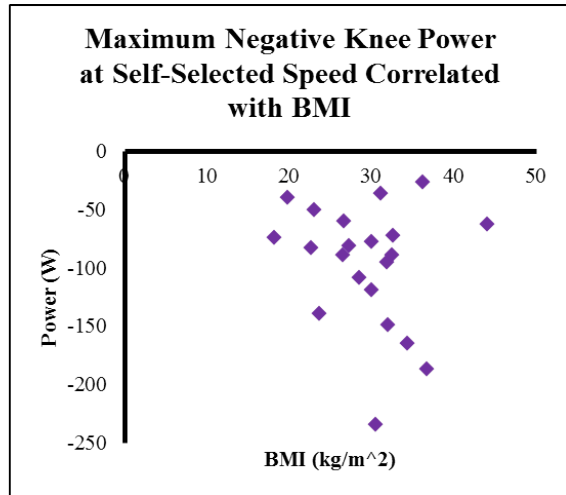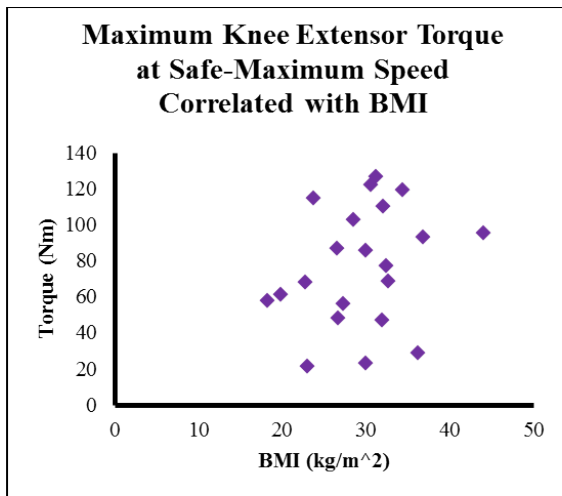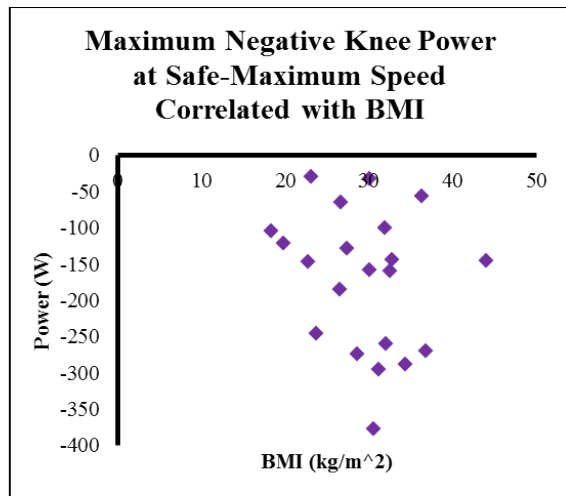Figure 23. Non-significant relationship between maximum negative knee power and BMI, r = 0.220, p > 0.05

Ankle Joint

We hypothesized that the ankle joint torques and powers would be directly correlated to BMI in a linear fashion throughout the entire range of BMI. Significant correlations were found at the ankle joint between maximum plantarflexor torque and BMI as well as between maximum positive ankle power and BMI at the self-selected normal speed. These data are presented in Figures 24 and 25.

Maximum Ankle Plantarflexor Torque at Self-Selected Speed Correlated with BMI

$y = 4.985x - 19.366$
$R^2 = 0.5914$

Figure 24. Significant relationship between maximum peak ankle plantarflexor torque and BMI, $r = .769$, $p < 0.05$

Maximum Positive Ankle Power at Self-Selected Speed Correlated with BMI

$y = 10.864x - 25.629$
$R^2 = 0.5847$

Figure 25. Significant relationship between maximum positive ankle power and BMI, $r = 0.765$, $p < 0.05$

Additional significant correlations were found between maximum peak ankle plantarflexor torque and BMI as well as maximum positive ankle power and BMI at the safe-maximum speed, seen in Figures 26 and 27.
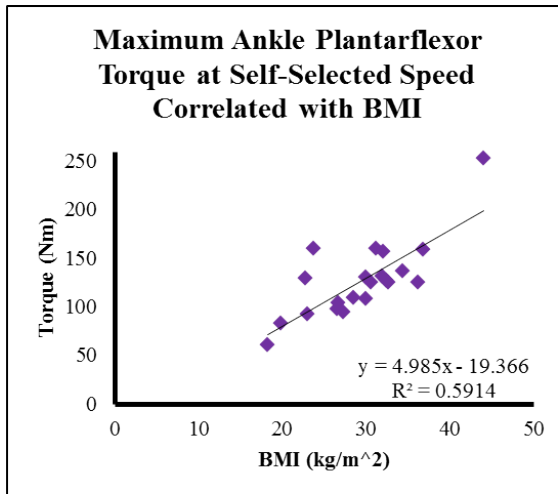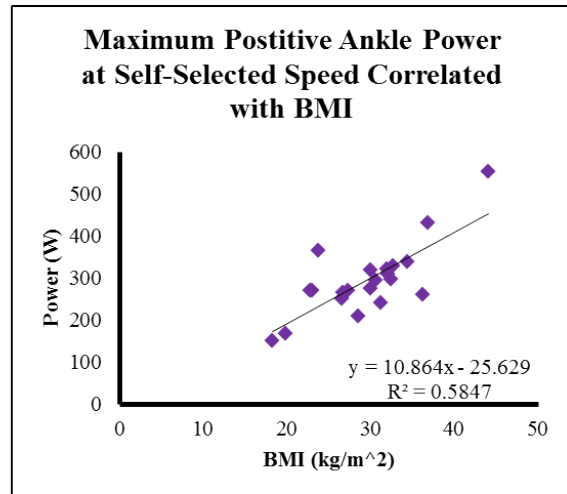
48
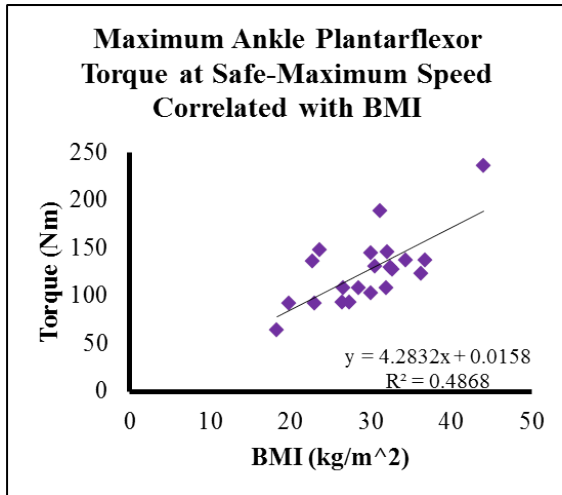
Figure 26. Significant relationship between maximum peak ankle plantarflexor torque and BMI, r = .698, p < 0.05
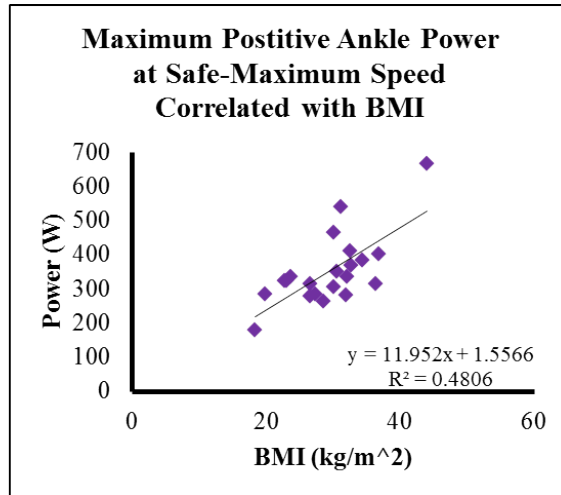
Figure 27. Significant relationship between maximum positive ankle power and BMI, r = 0.693, p < 0.05

Correlations of maximum ankle plantarflexor torque and BMI as well as the correaltions of maximum ankle positive power and BMI at both speeds were further evaluated using mass normalization. All correlations were found to be non-significant. These values are presented in Table 4. The lack of a significant relationship after normalizing for mass suggests that the relationships between ankle torque and BMI and ankle power and BMI are mass driven.

	Correlation with BMI at Self-Selected Speed	Correlation with BMI at Safe-Maximum Speed
Maximum Ankle Plantarflexor Torque	0.230	0.192
Maximum Ankle Positive Power	0.411	0.341

Table 5. Mass Normalized Correlation of Ankle Torques and Powers with BMI, *p<0.05

Ground Reaction Force

Similar to ankle joint biomechanics, we expected to find a direct, linear relationship between BMI and ground reaction forces for the entire range of BMI. At the self-selected speed, first maximum and second maximum ground reaction forces were found to be significantly correlated with BMI. Both braking and propulsive forces were also found to be significantly correlated with BMI at the self-selected speed. These data are represented in Figures 28 through 31. At the self-selected speed, additional analysis of mass normalized correlations showed non-significant relationships between first maximum vertical ground reaction force, maximum braking force, and maximum propulsive force and BMI. These non-significant correlations indicate the relationships between ground reaction forces and BMI are mass driven. A significant correlation was found between second maximum vertical ground reaction force in the opposite direction with r=0.641. This change in direction of correlation at a significant level confirms the correlation between second maximum vertical ground reaction force and BMI at the self-selected speed is mass driven.

First Maximum Vertical Ground Reaction Force at Self-Selected Speed Correlated with BMI
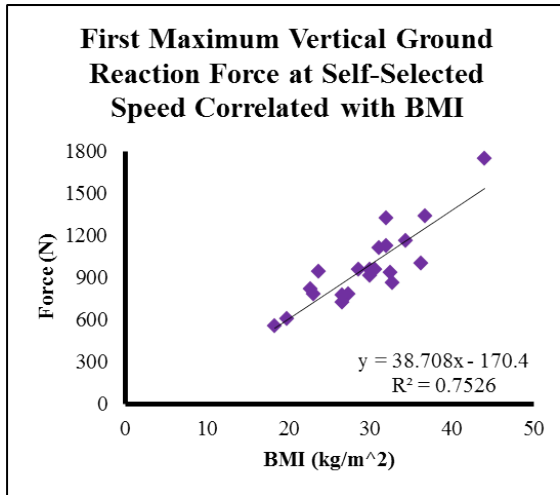
$y = 38.708x - 170.4$
$R^2 = 0.7526$

Figure 28. Significant relationship between first maximum vertical ground reaction force and BMI, $r = .868, p < 0.05$

Second Maximum Vertical Ground Reaction Force at Self-Selected Speed Correalted with BMI

$y = 29.853x + 34.459$
$R^2 = 0.7085$

Figure 29. Significant relationship between second maximum vertical ground reaction force and BMI, $r = 0.842, p < 0.05$

Maximum Braking Force at Self-Selected Speed Correlated with BMI

$y = -5.8607x - 28.067$
$R^2 = 0.437$

Figure 30. Significant relationship between maximum braking force and BMI, $r = .661, p < 0.05$

Maximum Propulsive Force at Self-Selected Speed Correlated with BMI

$y = 6.2156x + 11.875$
$R^2 = 0.5944$

Figure 31. Significant relationship between maximum propulsive force and BMI, $r = 0.771, p < 0.05$

First maximum and second maximum vertical ground reaction forces were both significantly correlated with BMI at the safe-maximum speed. Additionally, both maximum braking and maximum propulsive forces were significantly correlated with BMI at the safe-maximum speed. These data are depicted in Figures 32 through 35. Evaluation of these correlations was further explored by normalizing for mass. The results closely mimicked those at the self-selected speed. All correlations were non-

significant when normalized for mass with the exception of second maximum ground reaction force. This correlation was significant in the opposite direction with r= 0.721. These results indicate the correlations of ground reaction forces and BMI at the safe-maximum speed are mass driven relationships.

Figure 32. Significant relationship between first maximum vertical ground reaction force and BMI, r = .848, p < 0.05
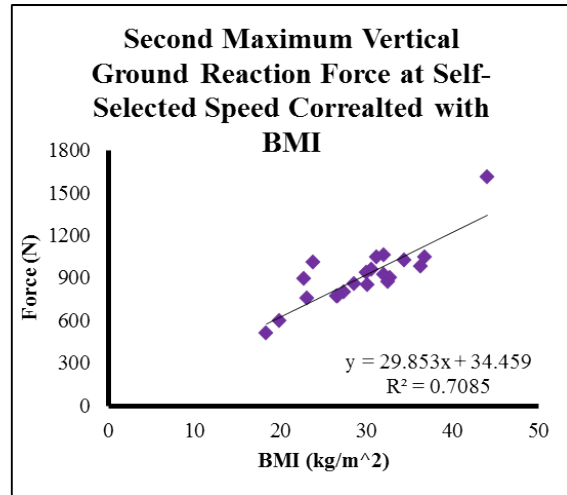

Figure 33. Significant relationship between second maximum vertical ground reaction force and BMI, r = 0.757, p < 0.05
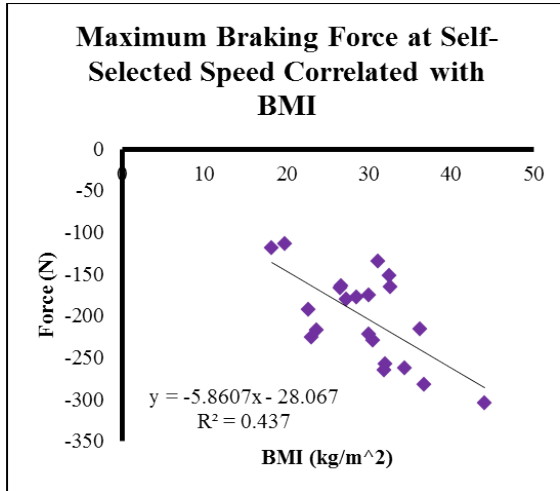

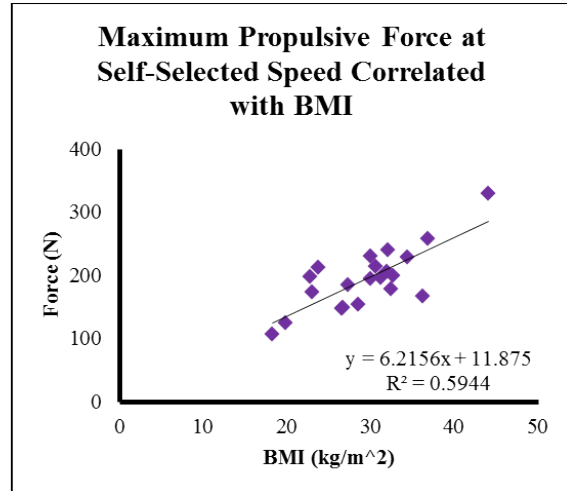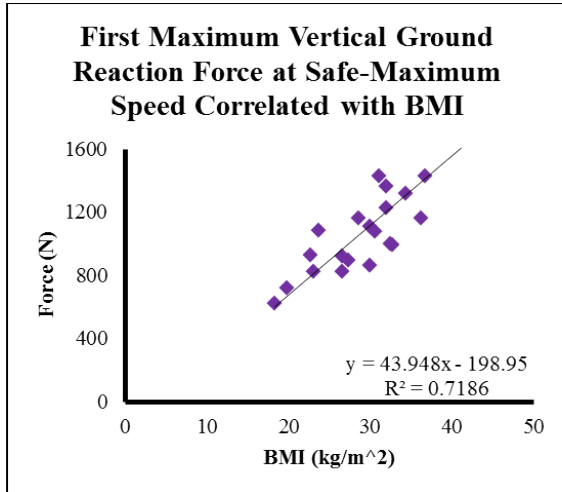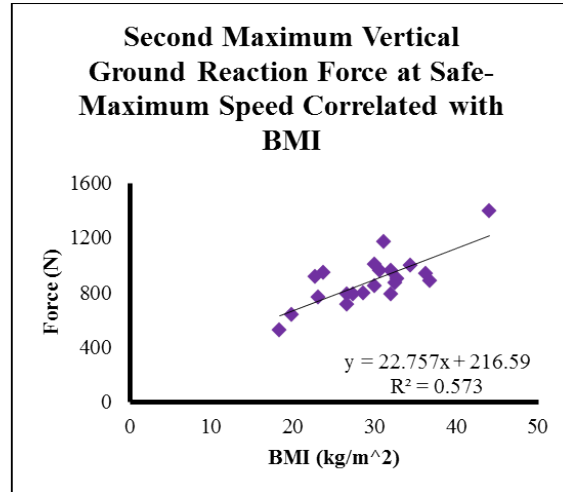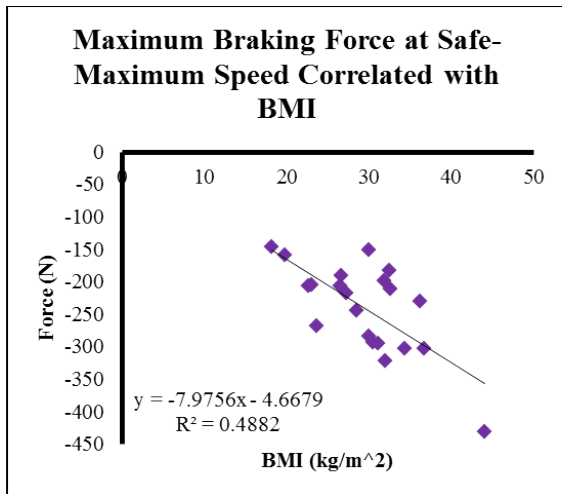Figure 34. Significant relationship between maximum braking force and BMI, r = .699, p < 0.05

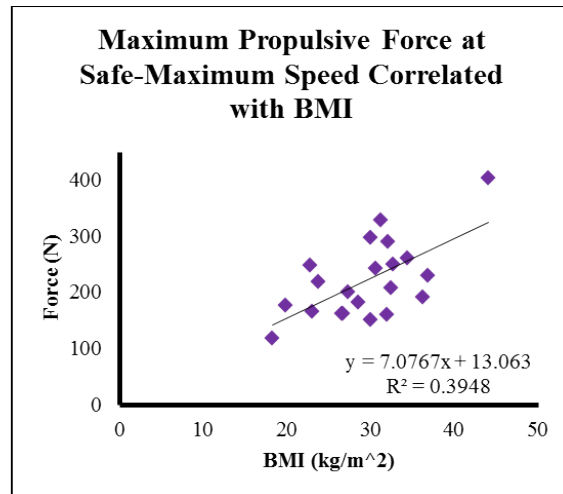Figure 35. Significant relationship between maximum propulsive force and BMI, r = 0.628, p < 0.05

We recognize that allowing a wide range of variability and age as well as not controlling for speed may have an effect on our results. Therefore, additional analysis of our correlations was performed to evaluate the effects of speed and age on joint torques and powers. Maximum hip extensor torque at the self-selected speed did prove to be affected by speed. Prior to evaluating the effect of speed, the correlation between maximum hip torque and BMI at the self-selected speed was $r = 0.458$, after adding speed to the analysis, our partial correlation became non-significant with $r = 0.327$. It was also found that age affected the correlation between maximum hip extensor torque and BMI at the self-selected speed, altering our initial correlation of $r = 0.458$ to $r = 0.304$. Very little effect of speed and age was found for our remaining variables. This is presented in Table 6. It can then be presumed that our correlations were, indeed, largely mass driven.

	Correlation with BMI at Self-Selected Speed	Correlation with BMI at Safe-Maximum Speed	Partial Correlation with Speed and BMI at Self-Selected Speed	Partial Correlation with Age and BMI at Self-Selected Speed	Partial Correlation with Speed and at Safe-Maximum Speed	Partial Correlation with Age and BMI at Safe Maximum Speed
Hip Maximum Extensor Torque	0.458*	0.629*	0.327	0.304	0.466*	0.476*
Hip Maximum Positive Power	0.769*	0.698*	0.106	0.108	0.126	0.180
Knee Maximum Extensor Torque	0.284	0.251	0.28	0.255	0.288	0.261
Knee Maximum Negative Power	-0.188	-0.220	-0.151	-0.156	-0.184	-0.232
Ankle Maximum Plantarflexor Torque	0.769*	0.698*	0.867*	0.738*	0.707	0.738*
Ankle Maximum Positive Power	0.765*	0.693*	0.73	0.668*	0.700*	0.743*
First Maximum Vertical Ground Reaction Force	0.868*	0.848*	0.895*	0.740*	0.786*	0.766*
Second Maximum Vertical Ground Reaction Force	0.842*	0.747*	0.938*	0.800*	0.749*	0.804*
Maximum Braking Force	0.661*	0.669*	-0.813*	-0.808*	-0.694*	-0.745*
Maximum Propulsive Force	0.771*	0.628*	0.787*	0.760*	0.675*	0.718*

Table 6. Correlations with Covariates of Speed and Age, *$p<0.05$

Summary

Due to an inadequate BMI range above 37 kg/m^2 within our sample, we were unable to fully evaluate the specific hypothesis of this research. We failed to identify a correlation between BMI and isokinetic knee flexor and extensor strength. It was found that a significant correlation existed between BMI and hip extensor torque and ankle plantarflexor torque at both the self-selected normal speed and the safe-maximum speed. However, we failed to find a correlation between BMI and knee extensor torque. No significant correlation was presented between BMI and maximum positive hip power or BMI and maximum negative knee power. Maximum positive ankle power was found to be significantly correlated with BMI. Significant correlations were also found between BMI and first maximum ground reaction force, second maximum ground reaction force, braking force, and propulsive force at both speeds.

Chapter 5: Discussion

This research was conducted to evaluate gait adaptations elicited by increasing magnitude of obesity. It was hypothesized that there is a critical magnitude of obesity, as measured with Body Mass Index, above which humans walk with an inverse relationship between BMI and hip and knee joint torque and power. The expected results were: that BMI will be: 1) directly related to GRFs and ankle joint torque and power and 2) directly related to hip and knee joint torques and powers from BMI 20 to ~35 kg/m^2 and indirectly related at higher BMI values. Such adaptations enable Class III obese individuals to modulate hip and knee torques and power to values seen in lean individuals while increasing torques and powers at the ankle despite having excessive body mass and ground reaction forces.

Our research methods were designed to asses kinetic characteristics among people ranging from lean (BMI = 18 kg/m^2) to obese (BMI = 44 kg/m^2). These characteristics were examined while participants ambulated across a level surface at a self-selected, normal pace (average 1.54m/s) and a safe-maximum pace (average 1.99 m/s). This chapter will discuss the results and related literature and hypothesis.

Development of the Hypothesis

As a direct relationship between mass and gravity, ground reaction forces (GRF) produced by a human during ambulation will increase as body mass increases. It can then be assumed that lower extremity joint torques, or the amount of force exerted by the muscles crossing the joint, will increase correspondingly. It could, similarly, be assumed that lower extremity joint powers exhibit a similar relationship to BMI. These ideas were

supported by Browning, et al., in 2007. This research showed that GRFs and hip, knee, and ankle joint torques increase linearly with BMI for an obese subject group with an average female BMI of 37.0 kg/m^2 and an average male BMI of 34.1 kg/m^2 (Browning, et al, 2007). Contradictory to these results, DeVita and Hortabagyi (2003) found joint torques and powers for Class III obese individuals to be statistically identical at the hip and knee joints but statistically higher at the ankle joint when compared to normal weight individuals. The mean BMI for this obese subject group was 42.3 kg/m^2 (DeVita et al., 2003). Additionally, Hortobagyi et al. (2011) showed no change in hip and knee torques during walking in people who changed from BMI of 43 kg/m^2 to 29 kg/m^2 after one year of weight loss. The difference between these studies led to our hypothesis that there is a critical magnitude of obesity, as measured by BMI, above which humans modulate hip and knee biomechanics so that they become inversely related to BMI. We expected this critical magnitude to occur between ~35 kg/m^2 and 45 kg/m^2.

Our results confirmed the direct relationship between ankle joint biomechanics and BMI as well as ground reaction forces and BMI. We were unable to fully test the hypothesis of hip and knee joint biomechanics due to a lack of subjects in the Class III obese range. We did find a significant relationship between hip joint torque and BMI at both walking speeds. We failed to identify a relationship between hip power or knee joint torques and powers and BMI at either walking speed.

20 Meter Walk Test and SF-36

Our results demonstrated a significant correlation between BMI and 20 Meter walk time. As participants' BMI increased, the time required to walk 20m also increased.

This contradicts the results of Browning and Kram (2005) who showed obese individuals walk at similar speeds as normal weight individuals. Despite the preferred walking speeds being similar, it was found that subjects generally prefer to walk at speeds at which their gross energy cost per distance is minimized (Browning and Kram, 2005). This may suggest that obese individuals have a similar gross energy cost per distance as normal weight individuals.

We failed to identify a significant relationship between the physical component scores (PCS) or mental component scores (MCS) on the SF-36 with BMI. These results differ slightly from those found by Wu et al. (2014) which demonstrated a correlation between abdominal obesity and physical component scores in five Chinese cities. Giuli et al. (2014) also reported a relationship between PCS and BMI in adults over the age of 60. It is possible that inaccurate self-assessment may be a factor in the lack of correlation between SF-36 and BMI. Additionally, our inability to identify a correlation between SF-36 scores and BMI may be a result of strict inclusion criteria for predominantly healthy individuals with the exception of increased BMI measures.

Isokinetic Strength

We were unable to identify a significant correlation between BMI and isokinetic knee flexor or extensor strength. This differs from the results found previously by Hulens, et al (2001). Their research showed greater absolute knee extensor and knee flexor strength in obese women; however, when corrected for fat free mass, obese women exhibited significantly lower knee extensor and flexor strength measures than their normal weight counterparts (Hulens, et al., 2001). Miyatake, et al. (2000) also reported

57

greater absolute leg strength in obese individuals age 20-60 years when compared to normal weight individuals age 20-60 years. This research implies obese individuals are functionally weaker than normal weight individuals. While subjects were encouraged to exhibit full muscle effort during each trial, it did appear that some subjects might not have fully exerted themselves. Our inability to identify a correlation may be a result of measurement error or subjects being unfamiliar or uncomfortable with the isokinetic dynamometer resulting in a less than maximal effort.

Correlation of Joint Torques and Powers with BMI

Our subject sample consisted of 21 predominantly healthy adults with BMIs ranging from 18 kg/m^2 to 44 kg/m^2. Only one subject included in the data set portrayed a BMI above 37 kg/m^2, resulting in a subject set more similar to that of Browning, et al. (2007). than DeVita and Hortobagyi (2003) or Hortobagyi et al. (2011). This lack of subjects with BMIs above 37 kg/m^2 made a full evaluation of our hypothesis unattainable.

The research did present statistically significant, linear relationships between BMI and first maximum and second maximum ground reaction forces as well as significant relationships between BMI and maximum braking and maximum propulsive forces at both speeds. These findings showed the relatively strong (explained variances were ~65% and 48% for vertical and A/P forces) and direct relationship between body mass index and the external ground reaction forces necessary to produce human walking. As mass increased, the overall forces applied in both vertical and anteroposterior directions also increased. These results are conceptually similar to those shown by Browning et al

58

(2007). The slopes of the prediction equations for the first maximum vertical force were 38 and 44 N/kg/m^2 at self-selected and maximum speeds indicating that for each unit increase in BMI the first maximum force increased approximately 41 N through this range of speeds. The mean change in maximum anteroposterior forces were approximately 7 N/kg/m^2. Browning et al. (2007) found that absolute maximum vertical GRF for obese individuals was approximately 58% greater than absolute maximum vertical GRF for normal-weight individuals at 1.75 m/s, a speed intermediate to our two speeds. When accounting for the difference in BMI in Browning et al (2007), their data showed a 39 N/kg/m^2 relationship between BMI and GRF which is very similar to our result. Maximum anterior-posterior GRF was found to be ~66% greater in obese individuals than normal-weight individuals at 1.75 m/s (Browning et al., 2007) which produced ~9 N/kg/m^2 relationship, again similar to our result.

Additionally, a statistically significant correlation was found between BMI and maximum ankle plantarflexor torque as well as between BMI and maximum positive ankle power at both speeds. These findings mimic the results of both Browning et al. (2007) and DeVita and Hortobagyi (2003). The slopes of the prediction equations for ankle torque were 5.0 Nm and 4.3 Nm at self-selected and maximum speeds suggesting that for each unit increase in BMI, ankle torque increased approximately 4.65 Nm through this range of speeds. Browning et al. (2007) found that at 1.75 m/s one unit increase of BMI produced an approximately 2.7 Nm increase in ankle torque. The differences between our results and Browning et al (2007) may be accounted for by the differences in BMI for the subject samples. The proposition that Class III obese individuals undergo gait adaptations with their increased body mass in which hip and

59

knee torques are modulated to imitate those of normal weight individuals while

compensating with ankle torque alterations is also supported by the results of Hortobagyi

et al. (2011). These results showed that with a weight loss of approximately 33.6%, no

changes in hip and knee torques were produced; however substantial decreases occurred

in ankle torque measurements.

The slopes of the prediction equations for ankle power were 10.9 W and 12.0 W

at self-selected and maximum speeds. These results suggest that for each unit increase in

BMI, ankle power increases approximately 11.45 W.

Maximum hip extensor torque was also found to be statistically correlated with

BMI. However the correlations between hip torque and BMI were weaker than the

correlations between maximum ankle plantarflexor torque and BMI at both speeds, as

presented in Table 5.

	Self-Selected Speed	Safe-Maximum Speed
Correlation of BMI and Hip Extensor Torque	0.458*	0.629*
Correlation of BMI and Ankle Plantarflexor Torque	0.769*	0.698*

Table 7. Correlation of BMI and Hip Extensor Torque and Ankle Plantarflexor Torque, *p<0.05

We failed to identify a significant correlation between maximum hip positive

power and BMI at either speed. No correlation was identified between maximum knee

extensor torque and BMI or maximum negative knee power and BMI for either speed.

These data showed weaker relationships between BMI and proximal hip and knee joint

biomechanics than BMI and distal ankle joint biomechanics. We could not fully test our

hypothesis because of the limited sample. The data do suggest however, that hip and knee

biomechanics may not be as tightly coupled to BMI and body mass as are ankle biomechanics. This observation agrees with DeVita and Hortobagyi (2003) and Hortobagyi et al (2011). Certainly further investigation into these relationships with more individuals in the Class III obesity group are necessary to fully identify the relationships among body mass and joint biomechanics during walking.

Our results varied from those found by Browning et al., when joint torques and powers were evaluated. Browning et al., showed a linear relationship between BMI and joint torques and powers at the hip, knee, and ankle. These increases in joint torques were attributed to using similar joint kinematics while experiencing higher GRF. Here, our results aligned more closely with the results found by DeVita and Hortobagyi in 2003. DeVita and Hortobagyi found similar joint torques and powers at the hip and knee of obese individuals and lean individuals when walking at the same velocity (DeVita et al., 2003). In fact, the results from DeVita and Hortobagyi found knee extensor torques of obese and lean individuals to be nearly equivalent (DeVita et al., 2003). Their sample also showed a reduction in knee power in the obese sample size (DeVita et al,. 2003). Additionally, an increase was seen in ankle torques and powers as BMI increased (DeVita et al., 2003).

Summary

The results from this research showed a linear increase in ankle joint torques and powers with increasing BMI. While we were unable to fully test our hypothesis due to recruitment difficulties, the lack of significant correlations between BMI and joint

torques and powers at the hip and knee suggest hip and knee biomechanics are less tightly coupled with BMI as are ankle joint biomechanics.

We recognize that our sample is highly skewed towards the female gender. The intent was to have an equal representation of both males and females. Our data may be biased by this excessive presence of female subjects. Therefore, our results may be more relatable to females. However, the variables tested, torque, power, and ground reaction forces, are basic components of human gait in all individuals. While research by Kerrigan et al., reported no significant differences in hip, knee, and ankle joint torques and powers between males and females, some differences may be seen in the Class III obese population (Kerrigan et al., 1998). More specifically, differences in the location of adipose tissue carried in males versus females, i.e. gynoid versus android obesity, may produce changes in gait biomechanics. It is our intention to continue subject recruitment and testing in order to provide a more gender-neutral subject pool as well as evaluate possible differences between males and females.

Overall, our correlations and Browning et al.'s comparisons directly support a relationship between BMI and ankle torques and powers as well as ground reaction forces. However, while Browning et al. show a direct relationship between hip and knee torques and powers with BMI using comparisons, our correlations do not support these relationships. Our evaluation of the correlations between our variables and BMI using mass normalization produced non-significant correlations. For our sample set, the relationship between BMI and mass was $r=0.885$. This strong correlation between BMI and mass and the lack of significant correlations between hip torque and ankle torque and power with BMI, when normalized for mass, demonstrate that mass is a driving factor in

62

the relationships. Additionally, Browning et al. demonstrated that when walking at 1.1 m/s, obese subjects exhibited the same absolute peak sagittal-plane knee net torque as normal weight individuals walking at a preferred typical speed of 1.4 m/s (Browning 2007). Our subject sample was instructed to walk at a normal pace as if attending work or an appointment. This resulted in an average normal pace of 1.54 m/s. Performing correlations with speed as a covariate confirmed our relationships between mass and joint powers and torques were not speed driven.

A more substantial sample size including more individuals in the Class III Obese range may show a level of magnitude of obesity at which behavioral gait adaptations occur to substantially alter hip and knee joint gait biomechanics. These adaptations may result in biomechanical changes implemented by Class III obese individuals to generate more efficient ambulation with their increased body mass.

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Appendix A

EAST CAROLINA UNIVERSITY
University & Medical Center Institutional Review Board Office
4N-70 Brody Medical Sciences Building· Mail Stop 682
600 Moye Boulevard · Greenville, NC 27834
Office **252-744-2914** · Fax **252-744-2284** · **www.ecu.edu/irb**

Notification of Initial Approval: Expedited

From: Biomedical IRB
To: Paul DeVita
CC:
      Paul DeVita
      Patrick Rider
Date: 6/24/2013
Re: UMCIRB 12-001701
      Gait Biomechanics in Obesity

I am pleased to inform you that your Expedited Application was approved. Approval of the study and any consent form(s) is for the period of 6/21/2013 to 6/20/2014. The research study is eligible for review under expedited category #4 and 6. The Chairperson (or designee) deemed this study no more than minimal risk.

Changes to this approved research may not be initiated without UMCIRB review except when necessary to eliminate an apparent immediate hazard to the participant. All unanticipated problems involving risks to participants and others must be promptly reported to the UMCIRB. The investigator must submit a continuing review/closure application to the UMCIRB prior to the date of study expiration. The Investigator must adhere to all reporting requirements for this study.

Approved consent documents with the IRB approval date stamped on the document should be used to consent participants (consent documents with the IRB approval date stamp are found under the Documents tab in the study workspace).

The approval includes the following items:

Name	Description
Announce & Daily Reflector Announcements.docx	Recruitment Documents/Scripts
Informed Consent - Obesity Continuum - Fall 2012.pdf	Consent Forms
Obesity Continuum Study Protocol - August 2012.pdf	Study Protocol or Grant Application
Phone Interview - Initial Health Survey - Obesity COntinuum - Fall 2012.pdf	Surveys and Questionnaires
SF36.pdf	Surveys and Questionnaires
SF36.pdf	Standardized/Non-Standardized Instruments/Measures

The Chairperson (or designee) does not have a potential for conflict of interest on this study.

IRB00000705 East Carolina U IRB #1 (Biomedical) IORG0000418
IRB00003781 East Carolina U IRB #2 (Behavioral/SS) IORG0000418

Appendix B

Health Survey (SF36)

http://www.sf-36.org/demos/sf-36.html

Today's

Date:_____

Name: Last:_____ First: _____ Date of Birth: _____

Please answer these questions by "check-marking" your choice. Please select only one choice for each item.

1- In general, would you say your health is:

   1. Excellent   2. Very good   3. Good   4. Fair   5. Poor

2- <u>Compared to ONE YEAR AGO</u>, how would you rate your health in general <u>NOW</u>?

   1. MUCH BETTER than one year ago.

   2. Somewhat BETTER now than one year ago.

   3. About the SAME as one year ago.

   4. Somewhat WORSE now than one year ago.

   5. MUCH WORSE now than one year ago.

3- The following items are about activities you might do during a typical day. **Does your health now limit you** in these activities? If so, how much?

Activities	1. Yes, Limited A Lot	2. Yes, Limited A Little	3. No, Not Limited At All
a) **Vigorous activities,** such as running, lifting heavy objects, participating in strenuous sports?	1. Yes, limited a lot	2. Yes, limited a little	3. No, not limited at all
b) **Moderate activities,** such as moving a table, pushing a vacuum cleaner, bowling, or playing golf?	1. Yes, limited a lot	2. Yes, limited a little	3. No, not limited at all

c) Lifting or carrying groceries?	1. Yes, limited a lot	2. Yes, limited a little	3. No, not limited at all
d) Climbing **several flights** of stairs?	1. Yes, limited a lot	2. Yes, limited a little	3. No, not limited at all
e) Climbing **one** flight of stairs?	1. Yes, limited a lot	2. Yes, limited a little	3. No, not limited at all
f) Bending, kneeing or stooping?	1. Yes, limited a lot	2. Yes, limited a little	3. No, not limited at all
g) Walking **more than a mile**?	1. Yes, limited a lot	2. Yes, limited a little	3. No, not limited at all
h) Walking **several** blocks?	1. Yes, limited a lot	2. Yes, limited a little	3. No, not limited at all
i) Walking **one** block?	1. Yes, limited a lot	2. Yes, limited a little	3. No, not limited at all
j) Bathing or dressing yourself?	1. Yes, limited a lot	2. Yes, limited a little	3. No, not limited at all

4- During the **past 4 weeks**, have you had any of the following problems with your work or other regular activities *as a result of your physical health*?

	Yes	No
a) Cut down on the **amount of time** you spent on work or other activities?	1. yes	2. No
b) **Accomplished less** than you would like?	1. yes	2. No
c) Were limited in the **kind** of work or other activities?	1. yes	2. No
d) Had **difficulty** performing the work or other activities (for example it took extra effort)?	1. yes	2. No

5. During the **past 4 weeks**, have you had any of the following problems with your work or other regular daily activities **as a result of any emotional problems** (such as feeling depressed or anxious)?

	Yes	No
a) Cut down on the **amount of time** you spent on work or other activities?	1. yes	2. No
b) **Accomplished less** than you would like?	1. yes	2. No
c) Didn't do work or other activities as **carefully** as usual?	1. yes	2. No

6. During the **past 4 weeks**, to what extent has your physical health or emotional problems interfered with your normal social activities with family, friends, neighbors, or groups?

1. Not at all   2. Slightly   3. Moderately   4. Quite a bit   5. Extremely

7. How much **bodily pain** have you had during the **past 4 weeks**?

1. None   2. Very mild   3. Mild   4. Moderate   5. Severe   6. Very severe

8. During the **past 4 weeks**, how much did **pain** interfere with your normal work (including both work outside the home and housework)?

1. Not at all   2. A little bit   3. Moderately   4. Quite a bit   5. Extremely

9. These questions are about how you feel and how things have been with you **during the past 4 weeks**. For each question, please give the one answer that comes closest to the way you have been feeling. How much of the time during the **past 4 week** …

	1. All of the time	2. Most of the time	3. A good bit of the time	4. Some of the time	5. A little of the time	6. None of the time
a) Did you feel full of pep?	1. All of the time	2. Most of the time	3. A good bit of the time	4. Some of the time	5. A little of the time	6. None of the time
b) Have you been a very nervous person?	1. All of the time	2. Most of the time	3. A good bit of the time	4. Some of the time	5. A little of the time	6. None of the time
c) Have you felt so down in the dumps that nothing could cheer you up?	1. All of the time	2. Most of the time	3. A good bit of the time	4. Some of the time	5. A little of the time	6. None of the time
d) Have you felt calm and peaceful?	1. All of the time	2. Most of the time	3. A good bit of the time	4. Some of the time	5. A little of the time	6. None of the time
e) Did you have a lot of energy?	1. All of the time	2. Most of the time	3. A good bit of the time	4. Some of the time	5. A little of the time	6. None of the time
f) Have you felt downhearted and blue?	1. All of the time	2. Most of the time	3. A good bit of the time	4. Some of the time	5. A little of the time	6. None of the time
g) Do you feel worn out?	1. All of the time	2. Most of the time	3. A good bit of the time	4. Some of the time	5. A little of the time	6. None of the time

h) Have you been a happy person?	1. All of the time	2. Most of the time	3. A good bit of the time	4. Some of the time	5. A little of the time	6. None of the time
i) Did you feel tired?	1. All of the time	2. Most of the time	3. A good bit of the time	4. Some of the time	5. A little of the time	6. None of the time

10. During the **past 4 weeks**, how much of the time has your **physical health or emotional problems** interfered with your social activities (like visiting with friends, relatives, etc.)?

    1. All of the time

    2. Most of the time.

    3. Some of the time

    4. A little of the time.

    5. None of the time.

11. How TRUE or FALSE is **each** of the following statements for you?

	1. Definitely true	2. Mostly true	3. Don't know	4. Mostly false	5. Definitely false
a) I seem to get sick a little easier than other people?	1. Definitely true	2. Mostly true	3. Don't know	4. Mostly false	5. Definitely false
b) I am as healthy as anybody I know?	1. Definitely true	2. Mostly true	3. Don't know	4. Mostly false	5. Definitely false
c) I expect my health to get worse?	1. Definitely true	2. Mostly true	3. Don't know	4. Mostly false	5. Definitely false
d) My health is excellent?	1. Definitely true	2. Mostly true	3. Don't know	4. Mostly false	5. Definitely false

Thank you!

76

Appendix C

TABLE 1. Information About SF-36 Health Status Scales and the Interpretation
of Low and High Scores

Concepts	No. of Items	No. of Levels	Meaning of Scores	
			Low	High
Physical functioning	10	21	Limited a lot in performing all physical activities including bathing or dressing	Performs all types of physical activities including the most vigorous without limitations due to health
Role limitations due to physical problems	4	5	Problems with work or other daily activities as a result of physical health	No problems with work or other daily activities as a result of physical health, past 4 weeks
Social Functioning	2	9	Extreme and frequent interference with normal social activities due to physical and emotional problems	Performs normal social activities without interference due to physical or emotional problems, past 4 weeks
Bodily pain	2	11	Very severe and extremely limiting pain	No pain or limitations due to pain, past 4 weeks
General mental health	5	26	Feelings of nervousness and depression all of the time	Feels peaceful, happy, and calm all of the time, past 4 weeks
Role limitations due to emotional problems	3	4	Problems with work or other daily activities as a result of emotional problems	No problems with work or other daily activities as a result of emotional problems, past 4 weeks
Vitality	4	21	Feels tired and worn out all of the time	Feels full of pep and energy all of the time, past 4 weeks
General health perceptions	5	21	Believes personal health is poor and likely to get worse	Believes personal health is excellent